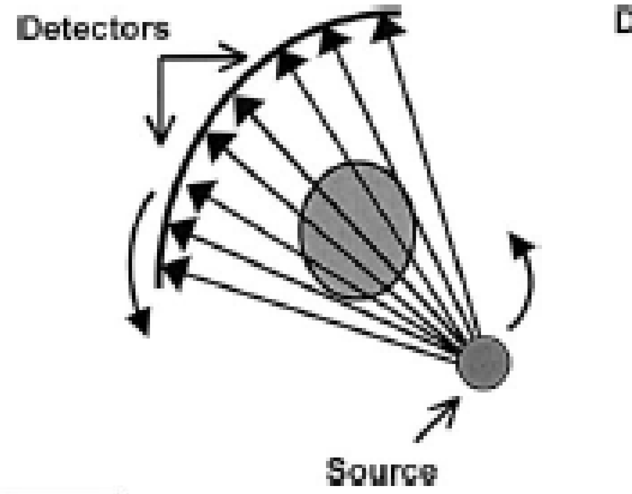
3rd generation components details

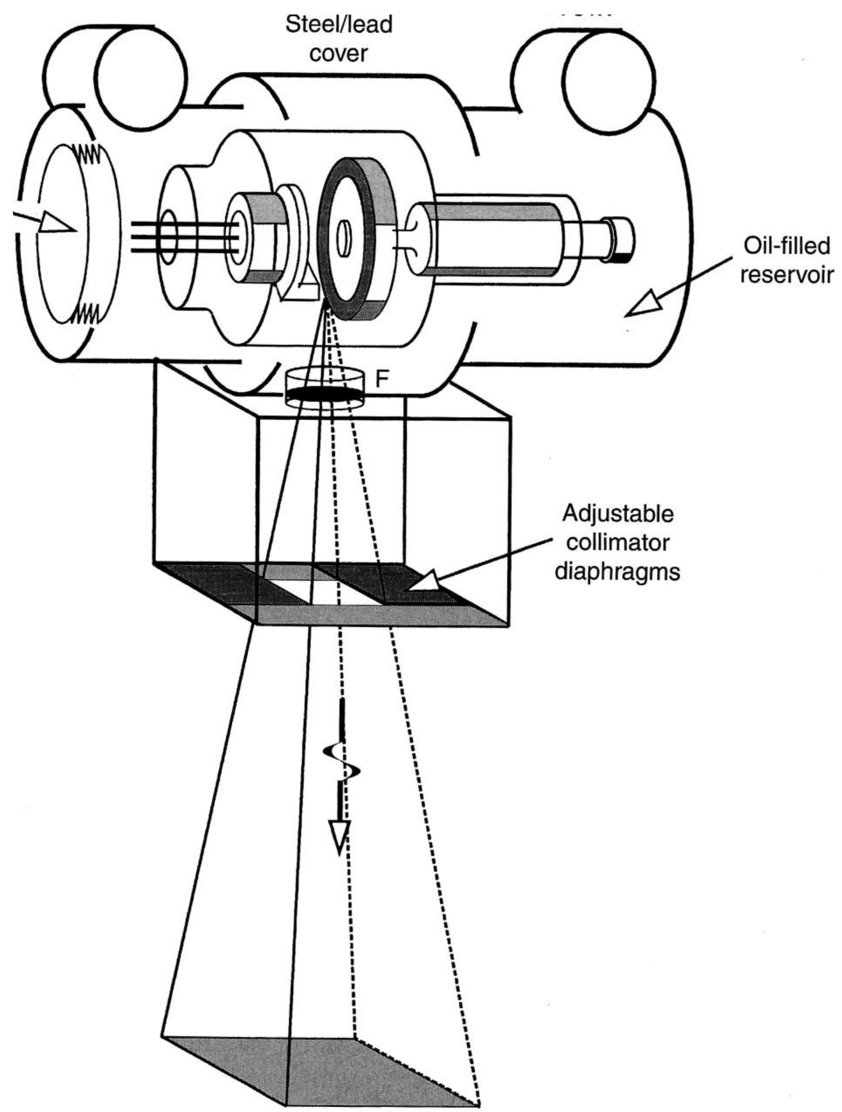
1)Rotating gantry :

- Tube and detectors are mounted on it
- rotation axis = z axis
- On helical scanners: rotation time = 1 s.
- Newest scanners : 0.3 s.
- Its possible to tilt the gantry up to 30^o about the vertical to tilt transaxial plane of the image e.g. brain scanning



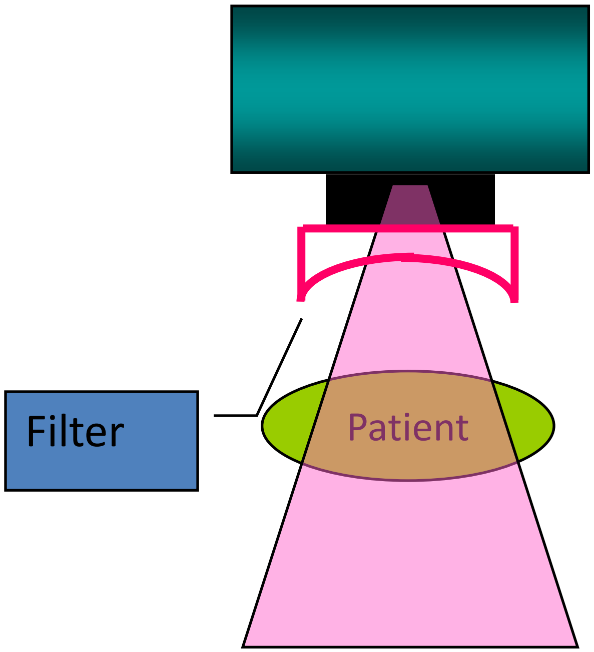
2) x-ray tube:

- Anode cathode axis is parallel to the axis of the scanner rotation → minimize anode heel effect
- Must be capable for prolonged exposure times at high mA (heat rating of 4MJ or more)
  - Achieved by
    - Having external cooling (oil heat exchanger)
    - Air within the gantry must be kept at low temperature
- Tube has two focal spot sizes (smallest = 0.6 mm)
- *kV range* of 80 - 140.
- *mA range* of 40 - 400



3) filters:

- 6 mm aluminum (additional copper was used in the past)
- protects patient from low-energy photons
- provides a beam closer to mono-energetic
- Different sizes of filters are used for head and body scanning



- Bowtie filters:
  - Used because the cross section of the patient is elliptical → noise at center is high , dose at periphery is high
  - This filter is Progressively thicker towards edges of the beam

  →even out transmitted intensities and beam hardening effect

4) Collimators:

- Restricts beam to slice of interest

i.e. X-ray beam is collimated to a wide fan beam sufficient to cover the patient’s cross section

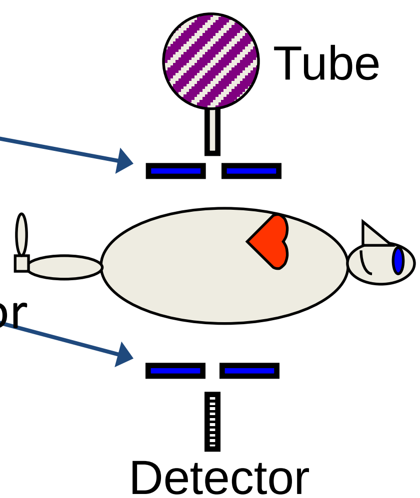
- Beam width in z axis = slice thickness (for single slice scanners) = 1 – 10 mm
- Two sets of collimators are used:

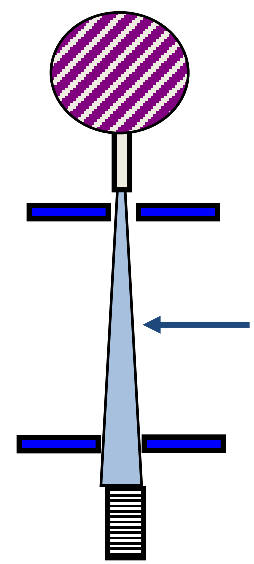
Pre-collimators

– between tube & patient

Post-collimators

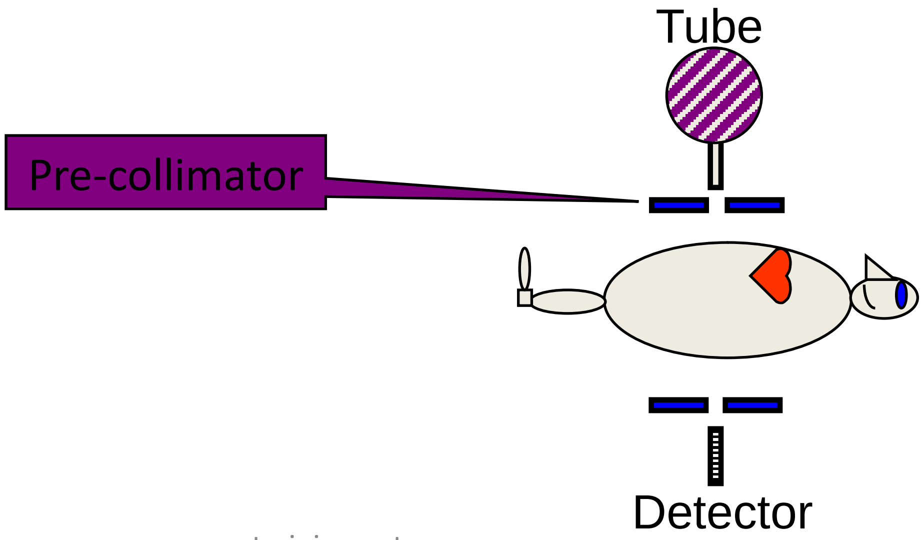
- between patient & detector





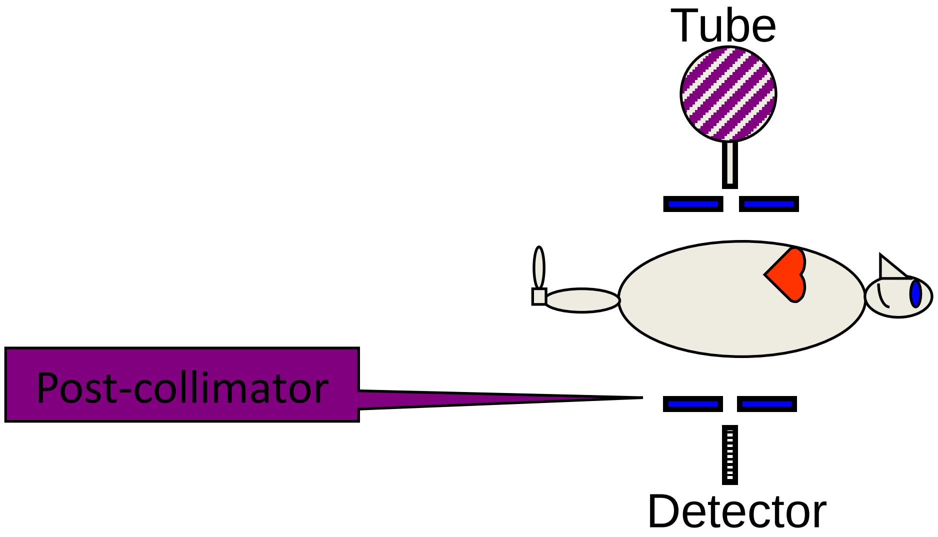
Pre-Collimation

- Constrains size of beam
- Reduces amount of scatter produced
- Designed to minimize beam divergence



Post-Collimation

- Used only for single slice scanners
- Mounted infront of the detectors
- Helps define slice (beam) thickness
- Reduces scatter radiation reaching detector



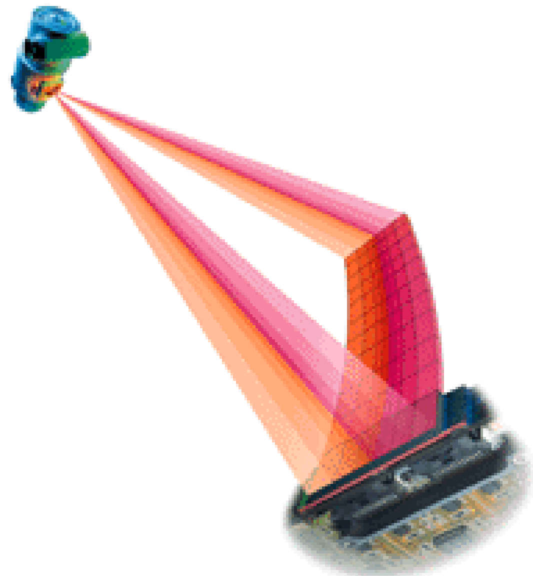
6) Detectors:

– Arranged in arc like array, with the Radius of the arc = focal distance (i.e. each detector is at the same distance from the source)

– Total number of detectors = 500-1000

– Characteristics of the ideal detectors used:

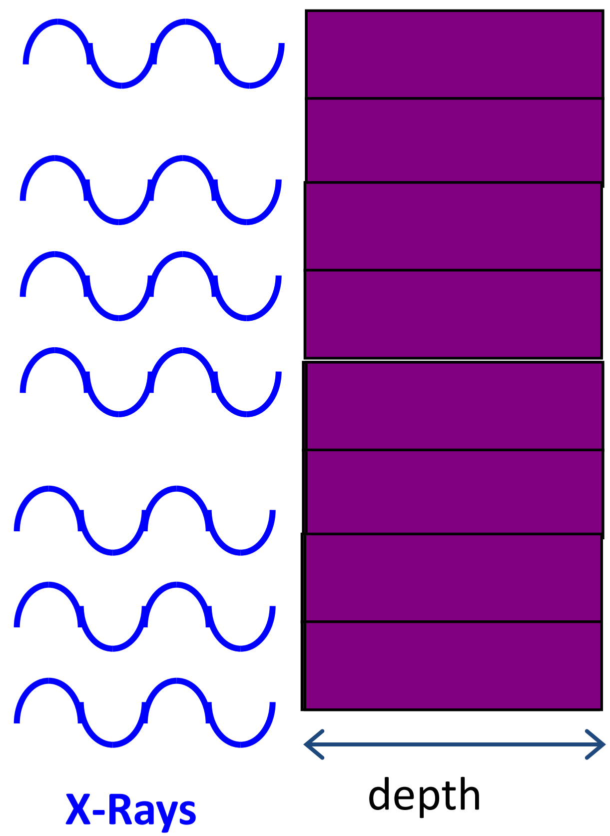
  - 1- small in size (to allow good spatial resolution)
  - 2- fast response and negligible afterglow (signal lag)
  - 3-stable (no need for repeated calibration)
  - 4- have high detection efficiency
  - 5-wide dynamic range
  - 6-noise free



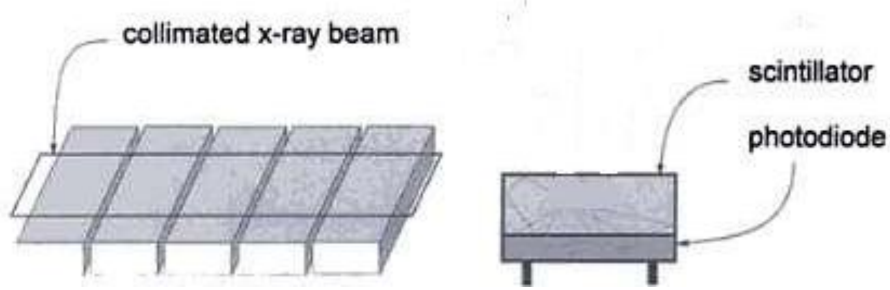
Two types of detectors used:

1- ionization chambers:

- Used in single slice scanners
- Detection efficiency = 60%
- Deep chambers are used (elongated in the direction parallel to the X-ray beam), This will:
  - 1- decrease the width → better spatial resolution
  - 2- increase the depth → increased sensitivity (absorption efficiency)
- Xenon gas is used:
  - 1- high Z (54) →increase detection efficiency
  - 2- kept under high pressure
    - more gas molecules encountered per unit path length →increase detection efficiency
- Not suitable for multislice scanners



2- solid state detectors:



- Composed of:
  - 1- Scintillator :
    - Cadmium tungstate
    - Bismuth germinate
    - Rare earth ceramic
  - Silicon photodiode
- Very high detection efficiency (up to 98%)
- Very small (1 x 0.5 mm^2)
- Stable with negligible afterglow

N.B: detection efficiency may decrease to 80% due to spacing (to prevent light crossover)

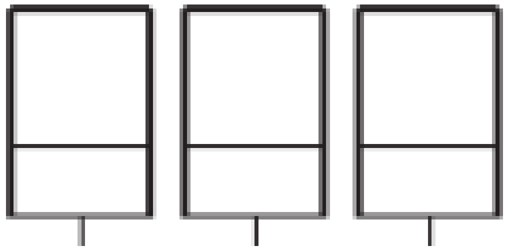
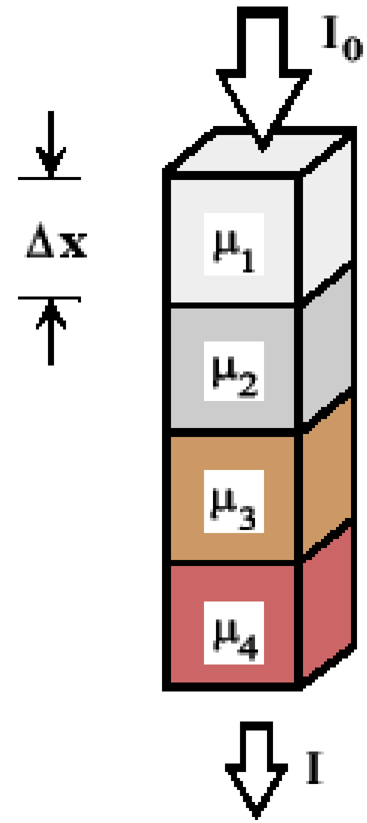


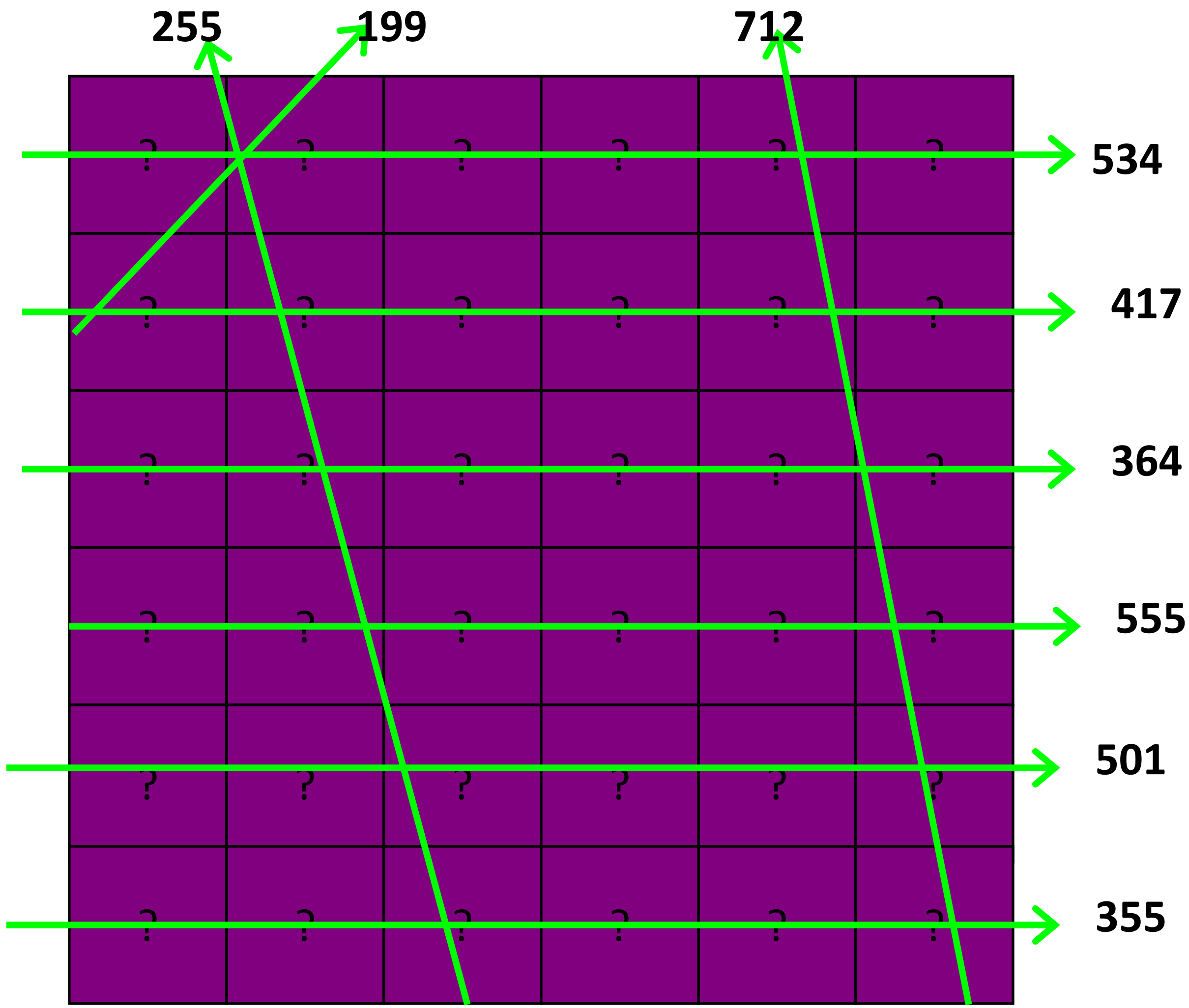
Image reconstruction

CT reconstruction problem

- Rays transmitted through multiple pixels during the scan
- Signal measured by each detector depends on
  - the attenuation coefficient of the voxels that are traversed by the beam
  - fraction of volume of the voxel through which the beam passes



$$I = I_o\, e^{-(\mu 1+\mu 2+\mu 3+\mu 4)\,(x1+x2+x3+x4)}$$



- Many measurements taken during single section scanning depend on:

1-number of detectors

2-number of measurements that are taken during a full rotation (e.g. at 0.5^{o} interval)

Techniques of data reconstruction:

1- back-projection:

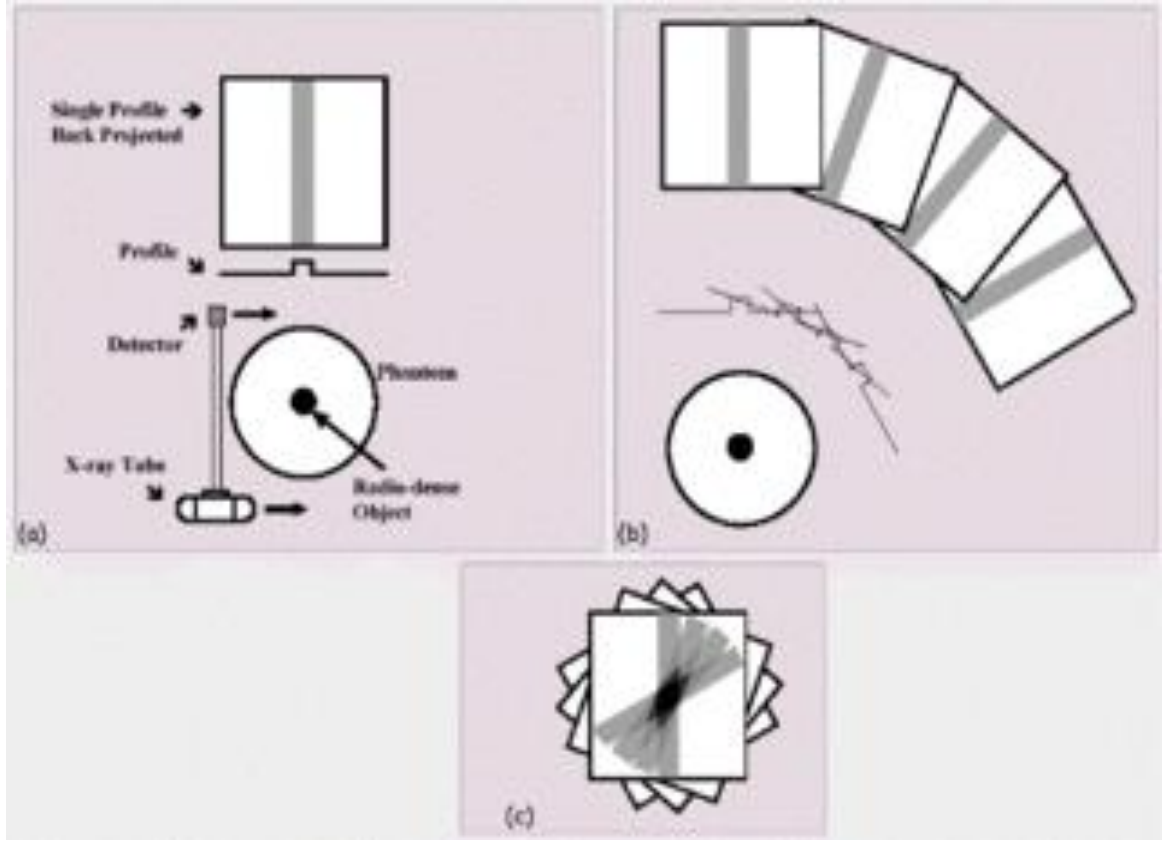
Done by considering that any attenuation of the beam has occurred uniformly along the path from the source to the detector

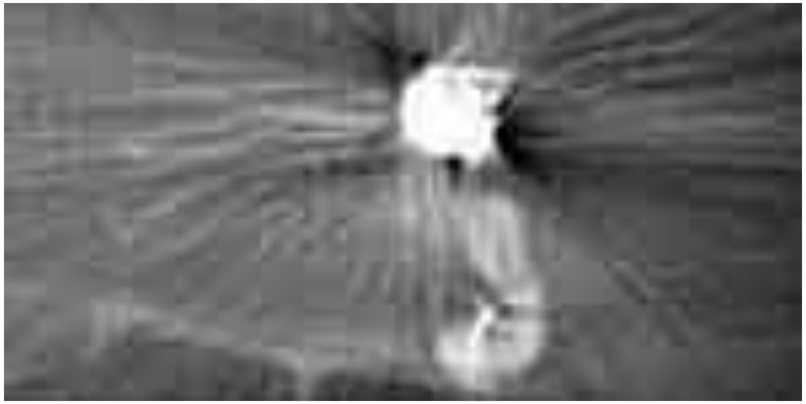
Example 1: The field contains a radio-dense object only

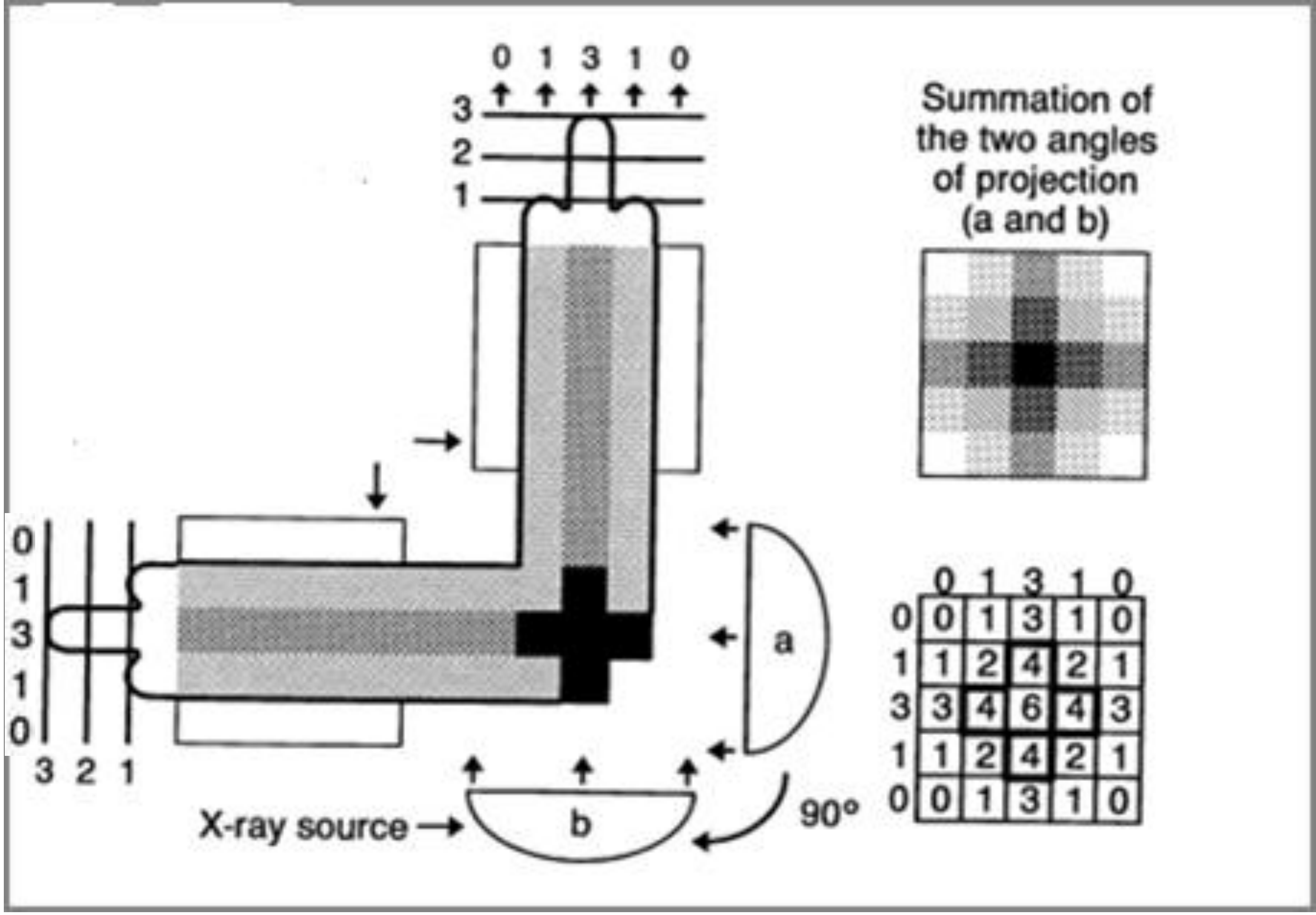
(a) Back-projection →dark stripe across the entire image plane

(b) Four profiles generated by scanning at slightly different angles around the phantom and are summed

(c) After back project an image of the radio-dense dot appears but surrounded by star like pattern





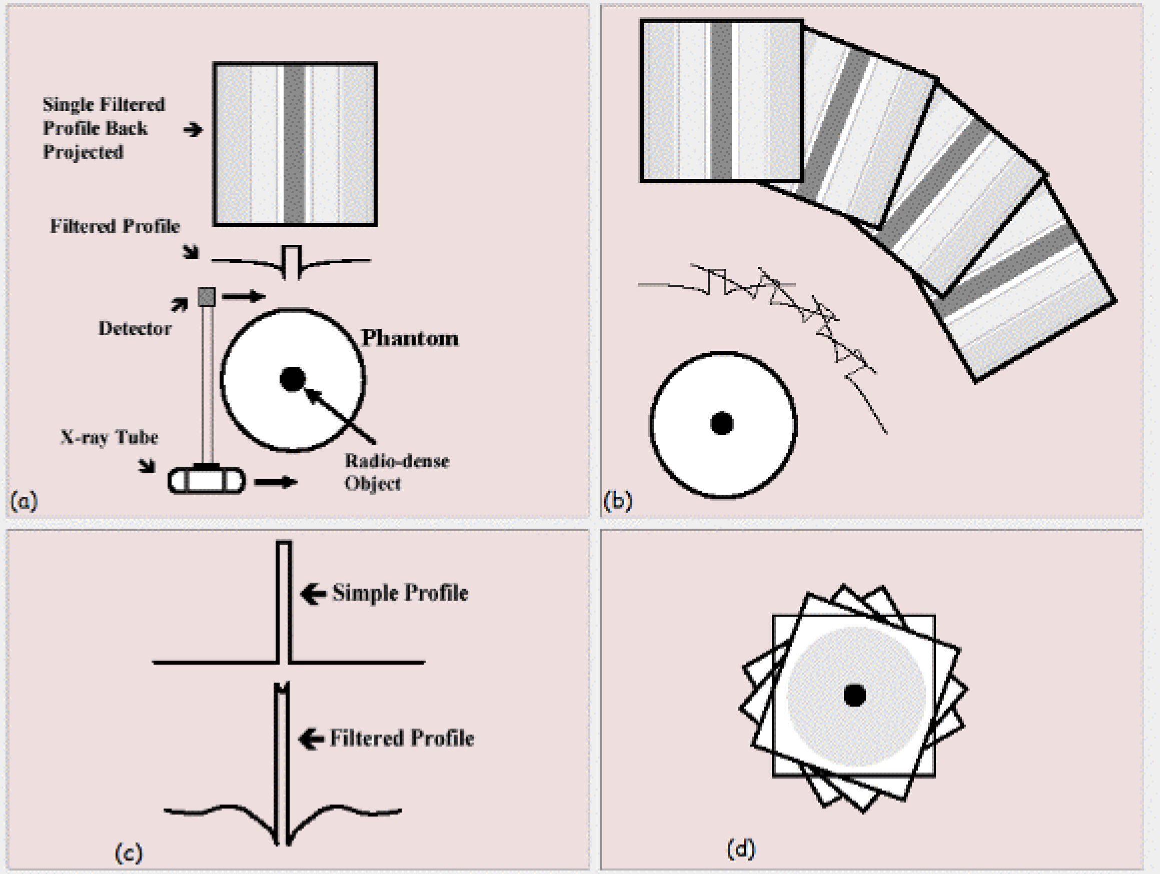


Example 2:

- The values of each angle is back projected
- The two back projected images are summed to give the final image representation
- The final image is a blurred version of the real image

2- filtered back-projection:

- Values that are projected-back use the data not only from the pencil beam itself , but also from the neighboring beams
- Data from the neighboring beams can be either subtracted or added to the projection to an extent that depend on the distance between the beams
- Overcome the problem of blurring
- Most common method used for reconstruction

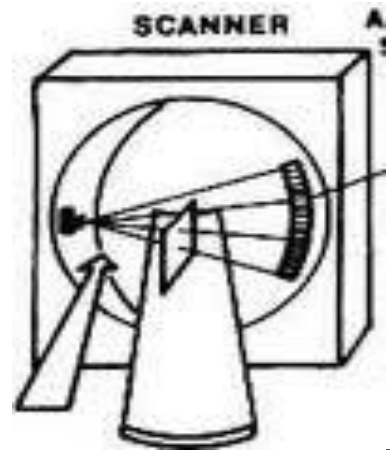


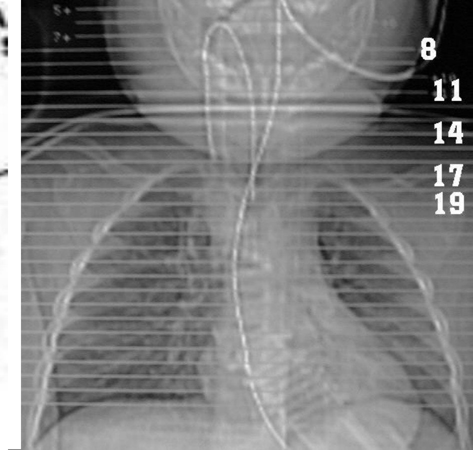
Notes:

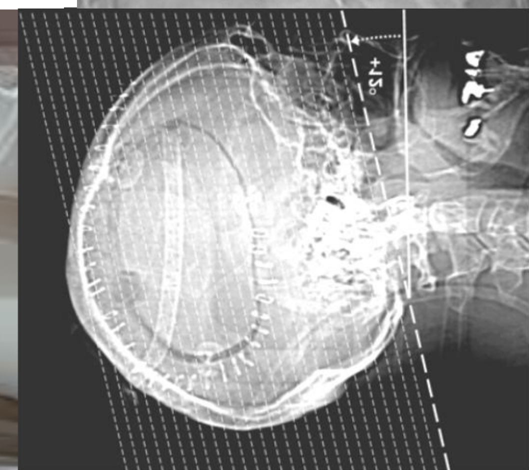
- Different back-projection filters are applied according to the imaging task (e.g. enhance bone resolution)
- It is possible to reconstruct from data collected 180^{o} only when rapid imaging process is required (e.g. CT fluoroscopy)

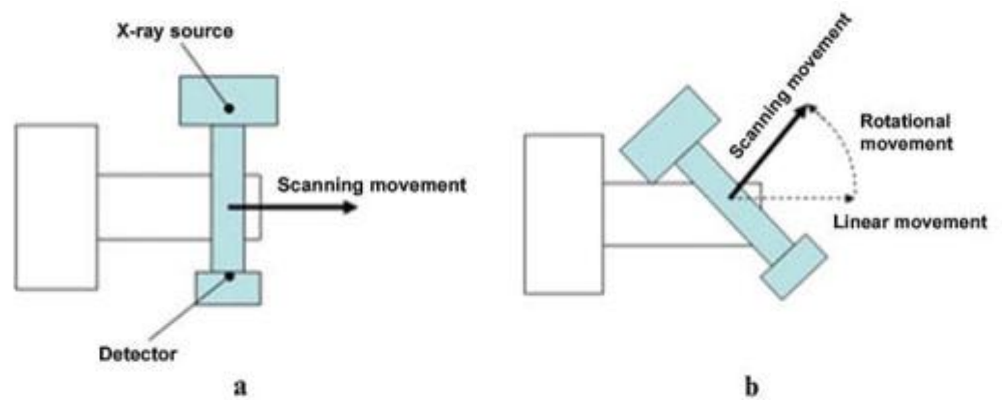
Scanned projection radiographs

- = scout view = scanogram = topogram
- Definition: Transmission image that are taken at fixed projection angle (antero-posterior or lateral)
- Process: collimator is set to the narrowest slice width → image is acquired as the table is moved through the gantry
- Disadvantage: poor spatial resolution
- Advantage: minimal scatter
- Uses: planning the CT sequence (scan start and end points , scan slice position)









CT fluoroscopy

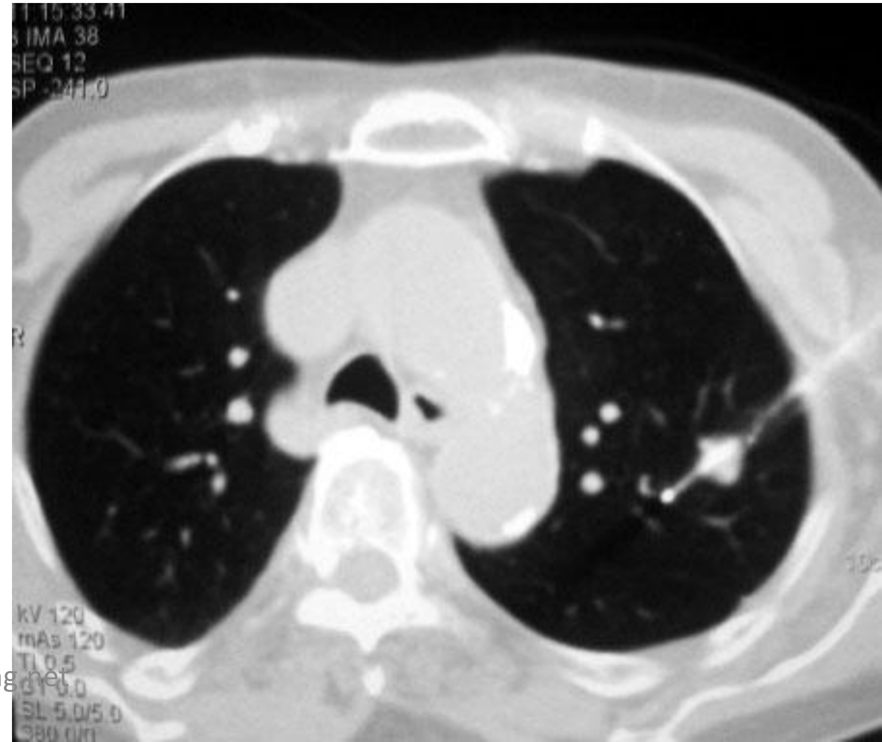
- Displaying CT image in real-time
- Method:

Continuous gantry rotation without table movement

Fast reconstruction techniques from 180^o data sets

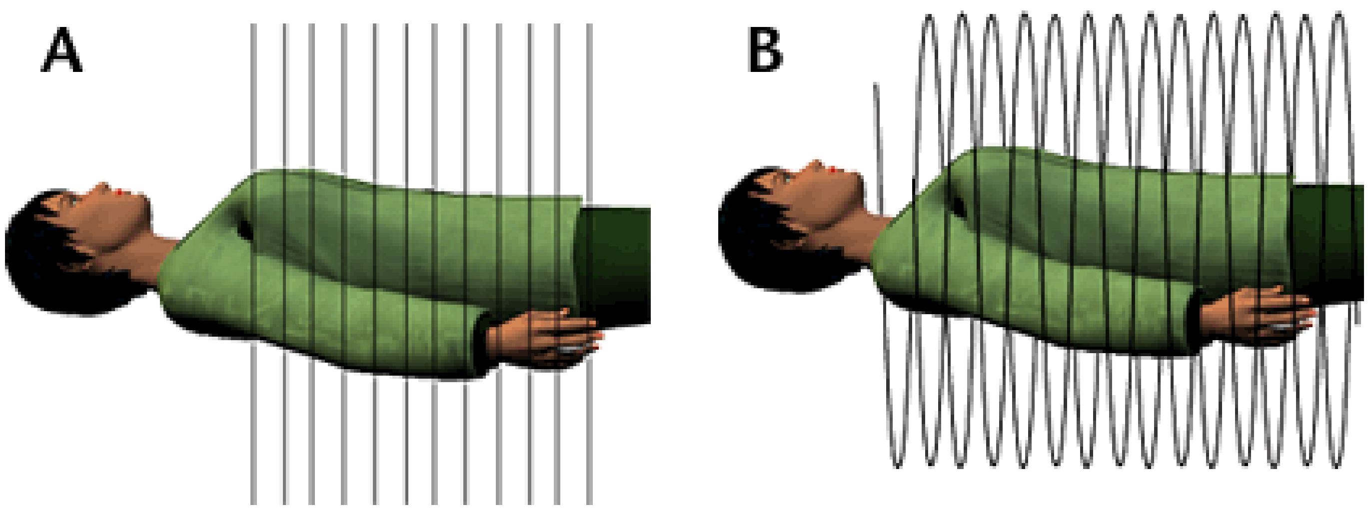
Achieve frame rate of 5 frames/s.

- Uses: CT guided biopsy
- Disadvantage: high patient's skin dose (scanning is confined to a narrow region of the body)

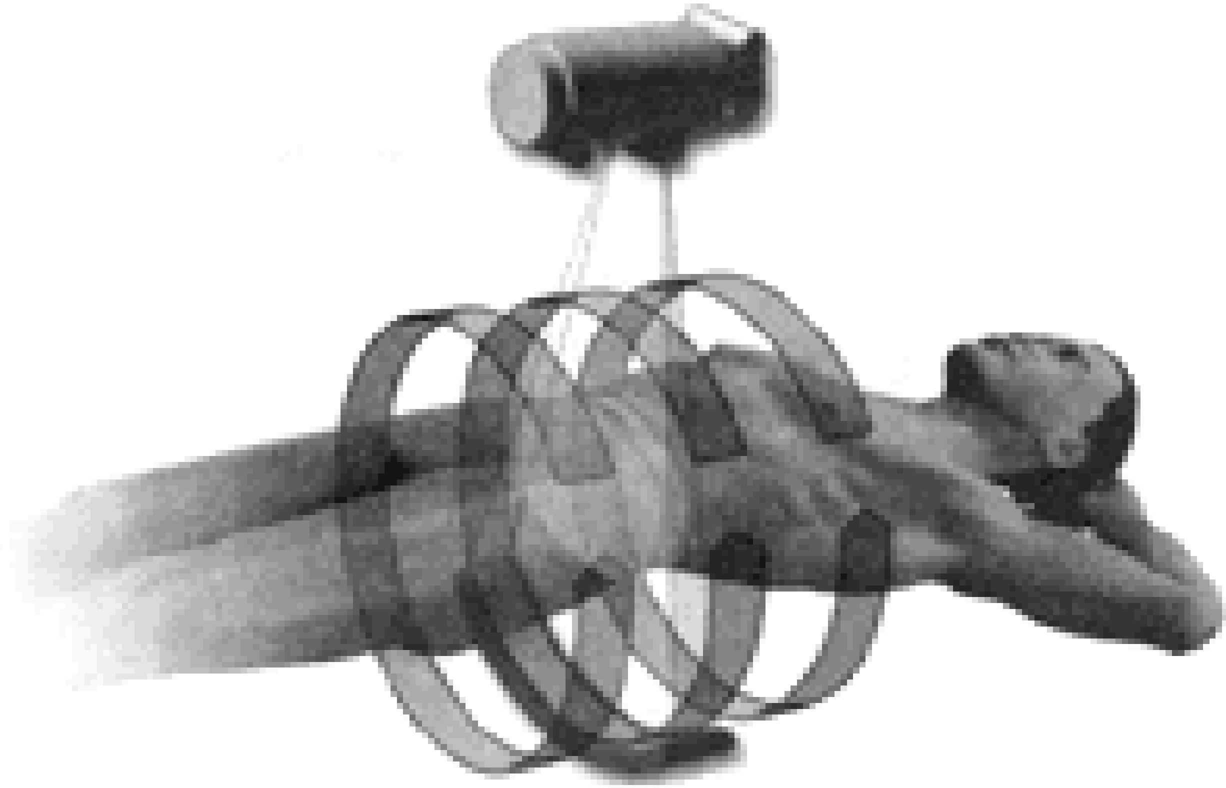


Conventional vs. Helical (spiral) scanning

CONVENTIONAL AND SPIRAL/HELICAL CT

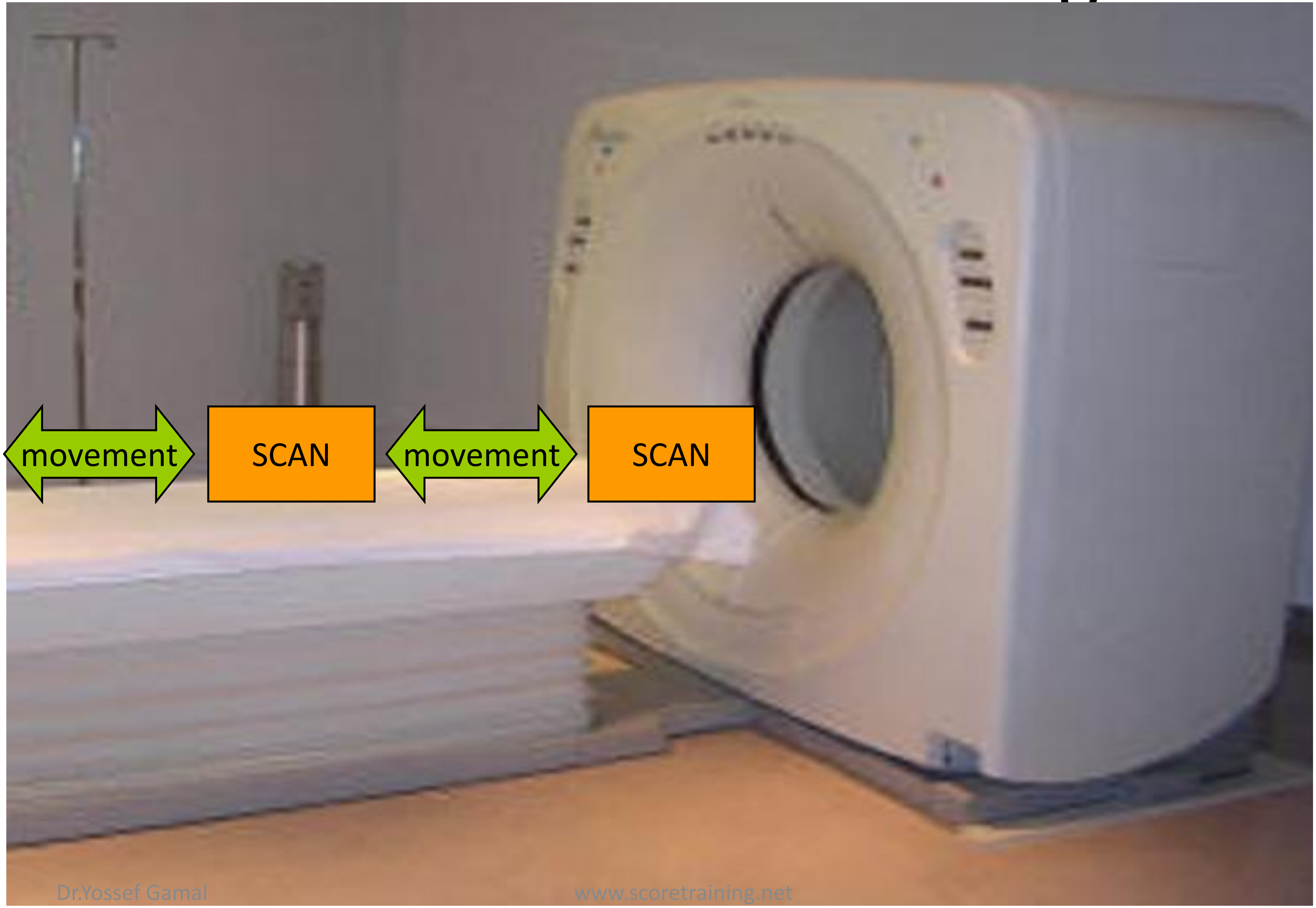


Conventional scanning
= step and shoot
= axial scanning
=slice-by slice data acquisition



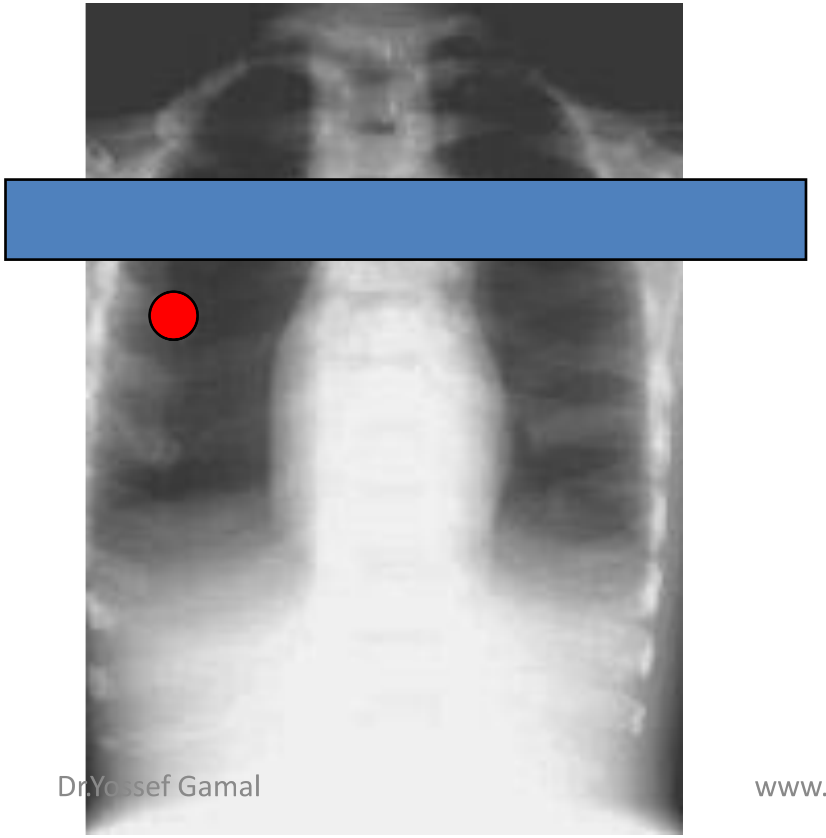
1) x-ray tube rotates around the patient to collect data from a single slice
2) table move for a certain distance along Z- axis of the scanner
3) Another slice is scanned

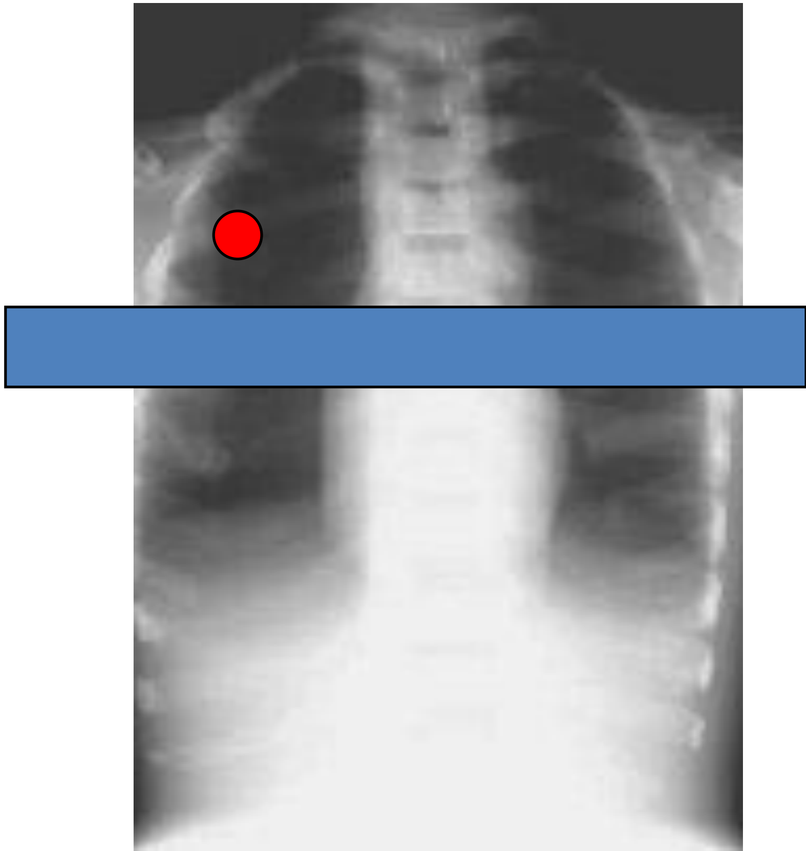
Conventional scanning



LIMITATIONS OF CONVENTIONAL CT

- **1- Longer exam time**
- **2- Slice misregistration: e.g. Patient's breathing**
- **3- Inaccurate generation of 3-d images**





4- maximum gantry rotation of 360^o

wires are needed to :

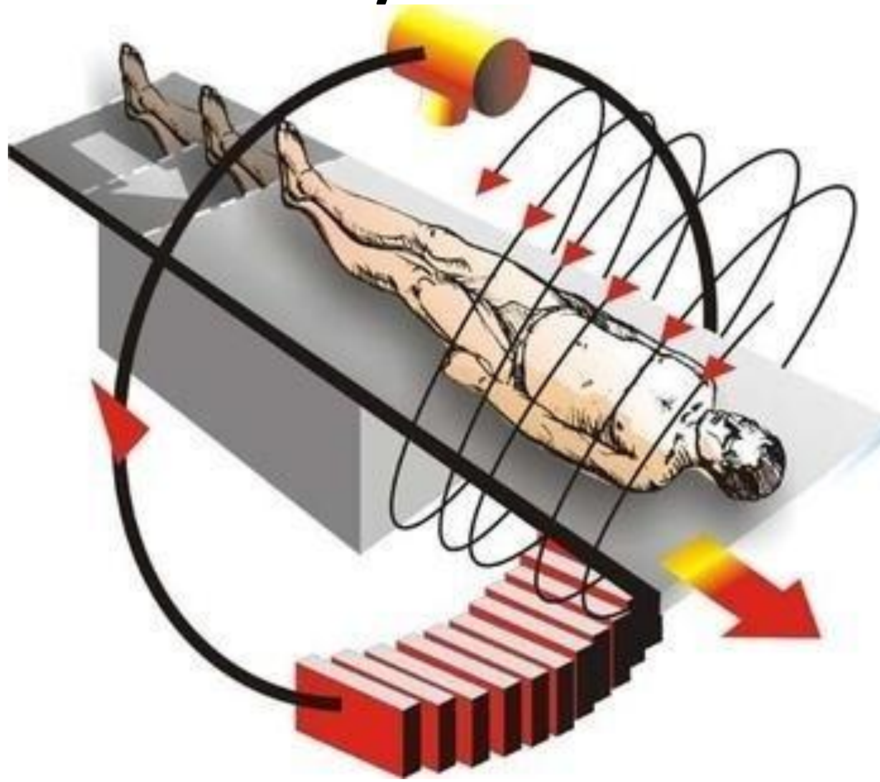
A- provide high voltage supply to the tube from the generator

B- sending signals from the detectors to the computer

→ the gantry was able to rotate for maximum angle of 360^o (without cable stretching)

→ following single rotation , the gantry return back to start position

SPIRAL/HELICAL CT

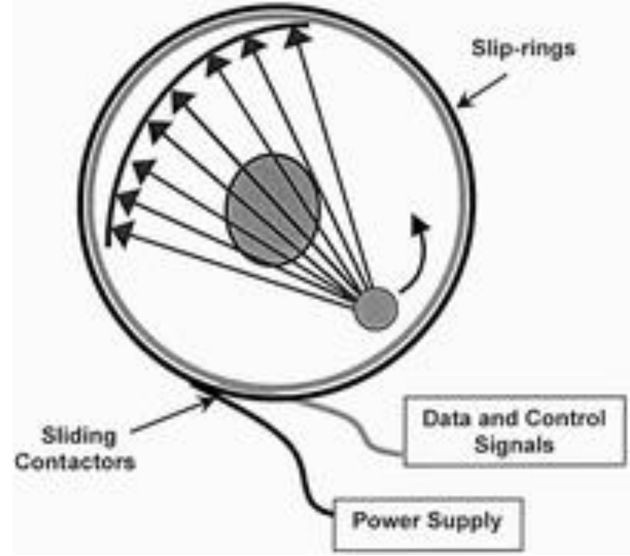


continuously rotating gantry with continuous longitudinal couch movement → Tube traces spiral path with respect to patient → volumetric imaging

slip ring technology

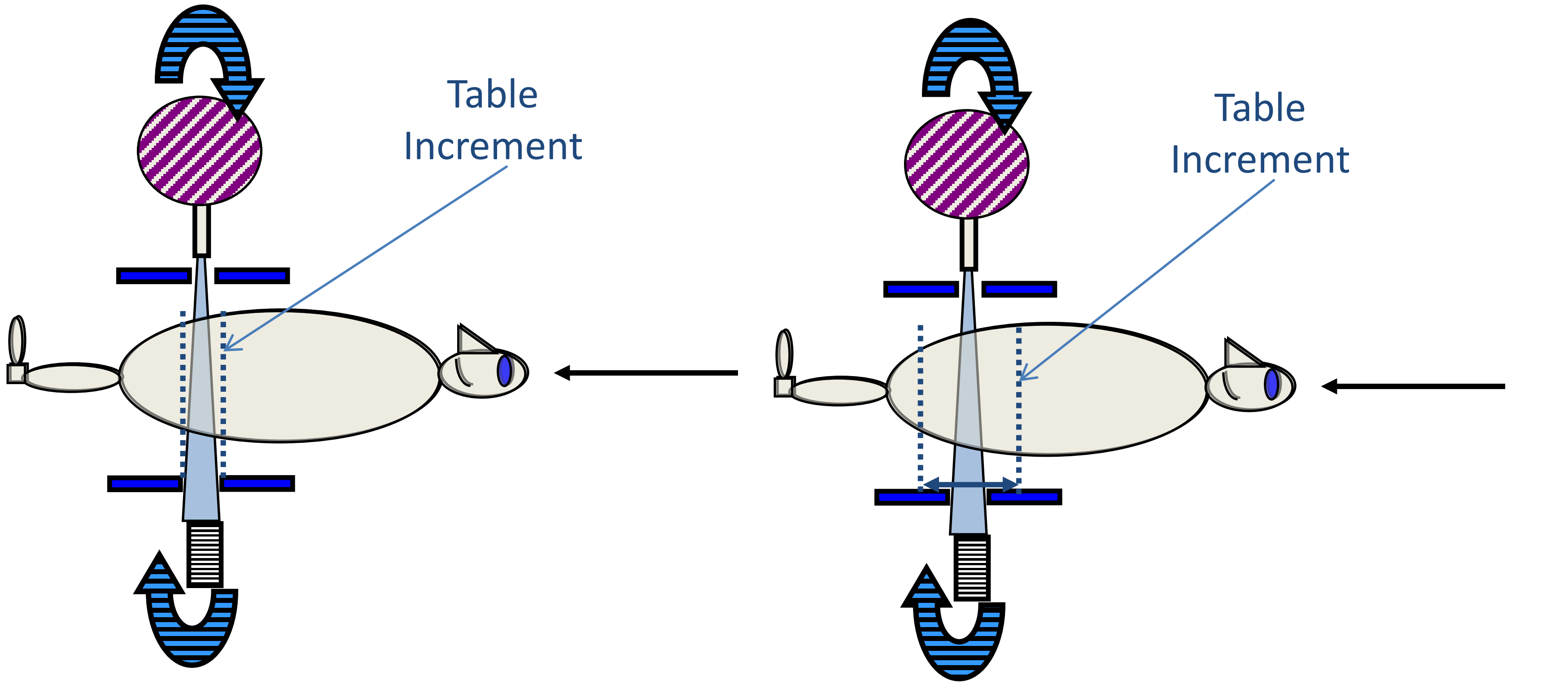
- **Function: induce Continuous gantry rotation in helical scanners**
- Metal ring is mounted on the gantry,
- good contact between the gantry and the ring is maintained as the gantry rotates using Connector
- Two ring connector system are used: one to take the detector signal and one to transfer tube voltage

→permit continuous gantry rotation



Helical scanning Pitch

$$\text{Slice Pitch} = \frac{\text{table motion during one rotation}}{\text{slice thickness}}$$

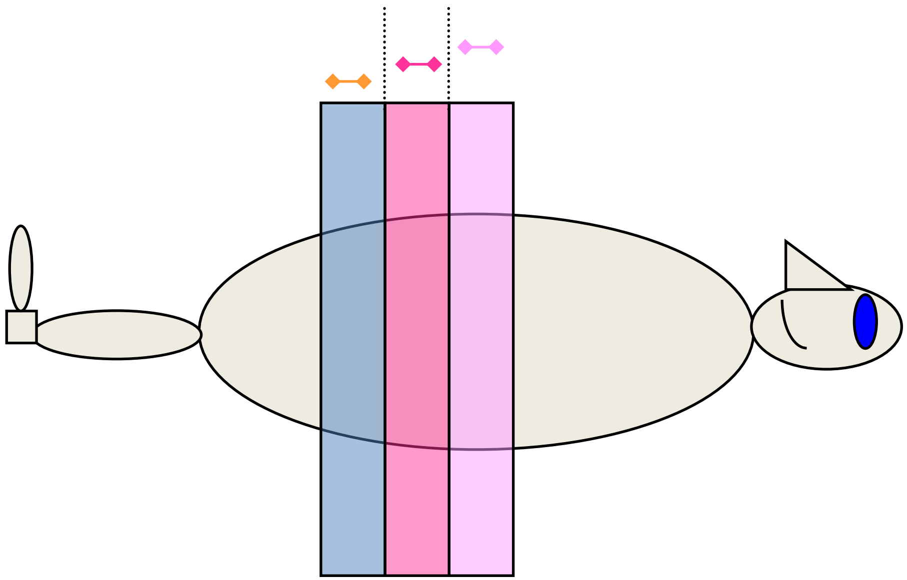


- N.B: Slice thickness in helical scanning is determined by collimation

Pitch = 1

- Pitch = 1 means slices abut one another
- Example: slice thickness = 10 mm, rotation time = 1 sec. , table speed = 10 mm/s

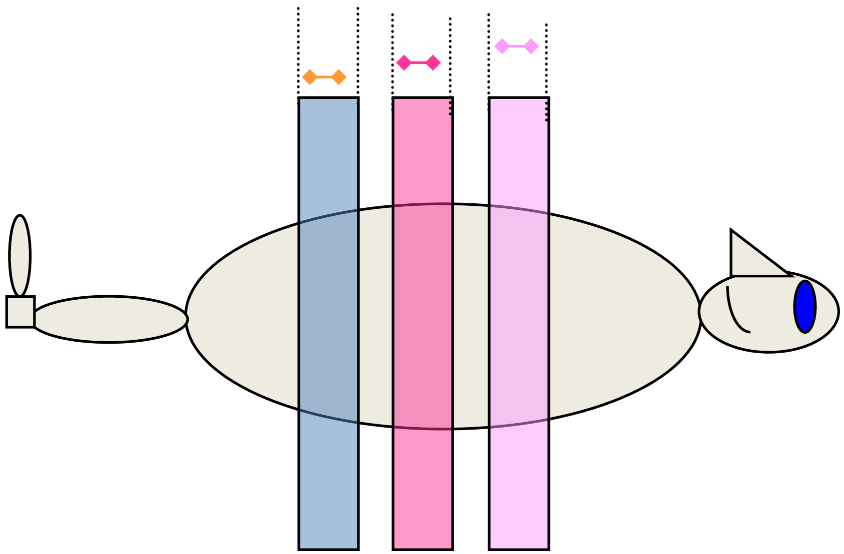
$$\text{Slice Pitch} = \frac{\text{table motion during one rotation}}{\text{slice thickness}}$$



Pitch >1

- Pitch > 1 means gap between slices

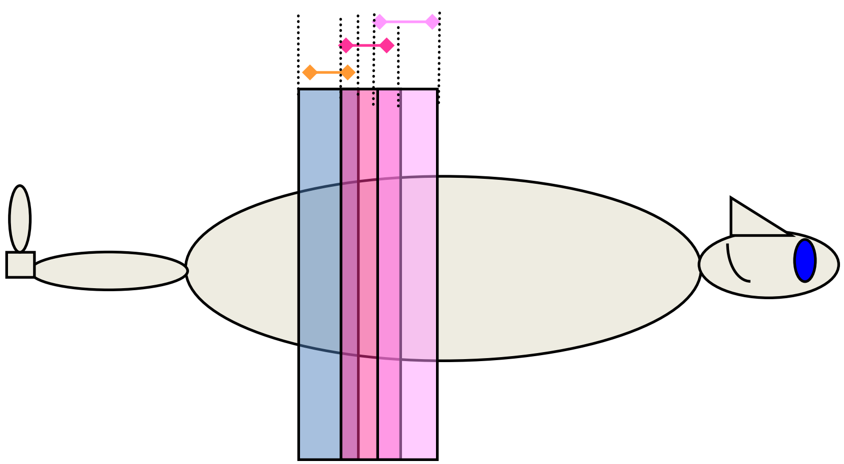
$$\text{Slice Pitch} = \frac{\text{table motion during one rotation}}{\text{slice thickness}}$$

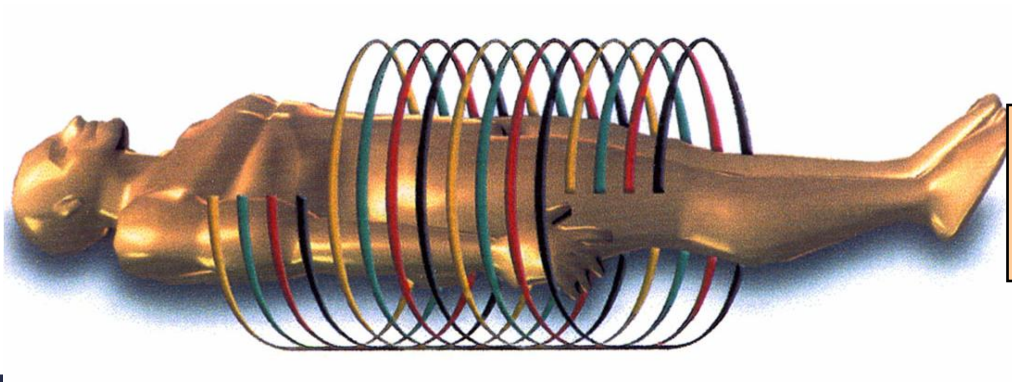


Pitch <1

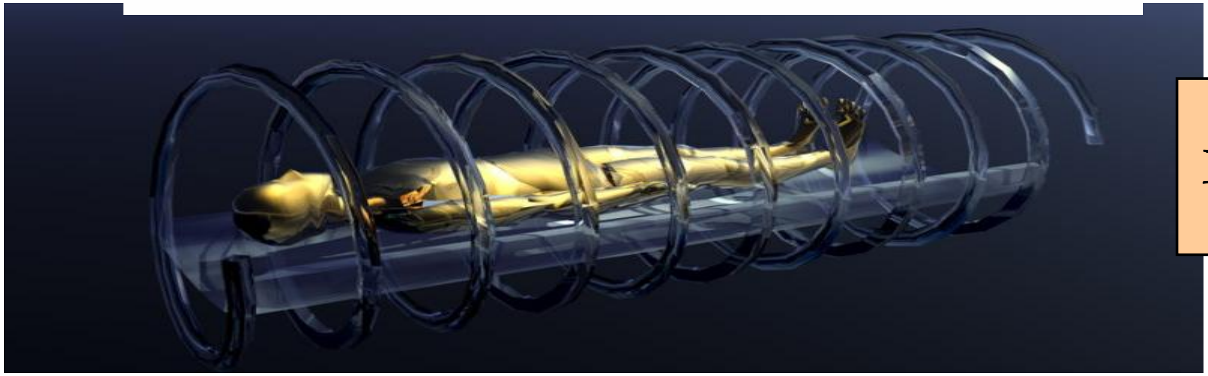
- Pitch < 1 means overlap in slices
  - Can improve visualization of objects

$$\text{Slice Pitch} = \frac{\text{table motion during one rotation}}{\text{slice thickness}}$$





SMALL PITCH

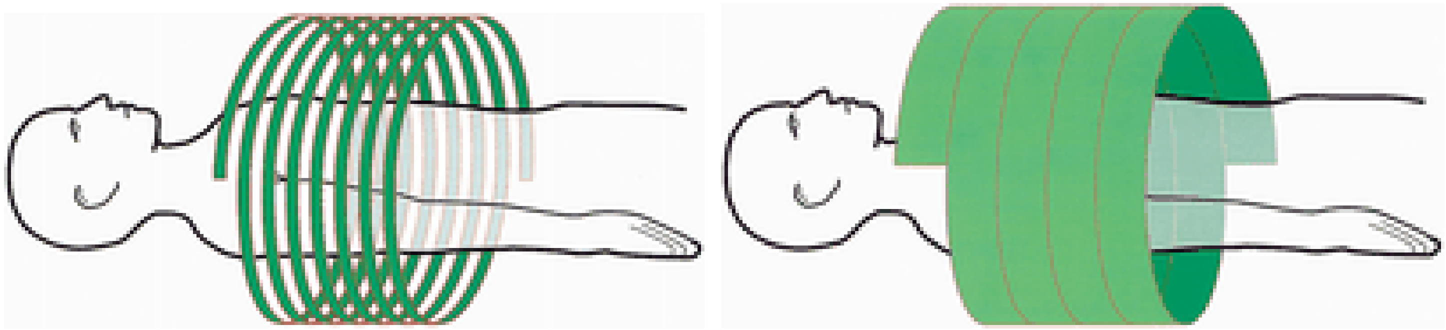


LARGE PITCH

PITCH ↑ SCAN TIME ↓

but PITCH ↑ IMAGE QUALITY ↓

Volume coverage

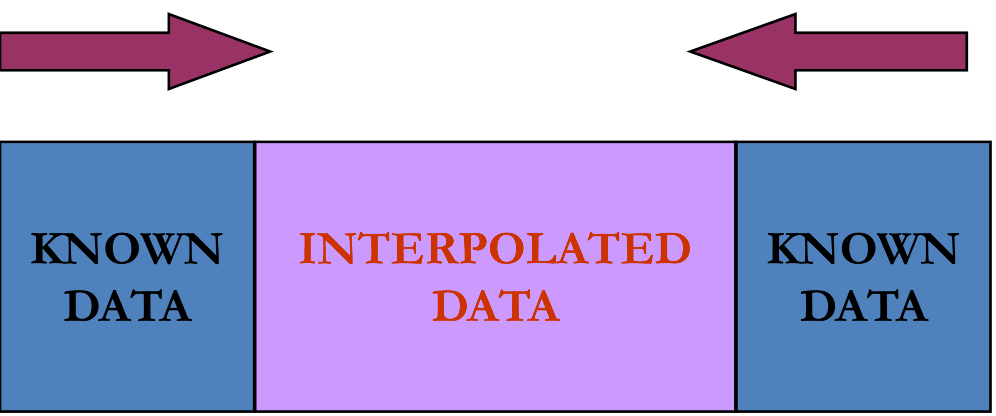


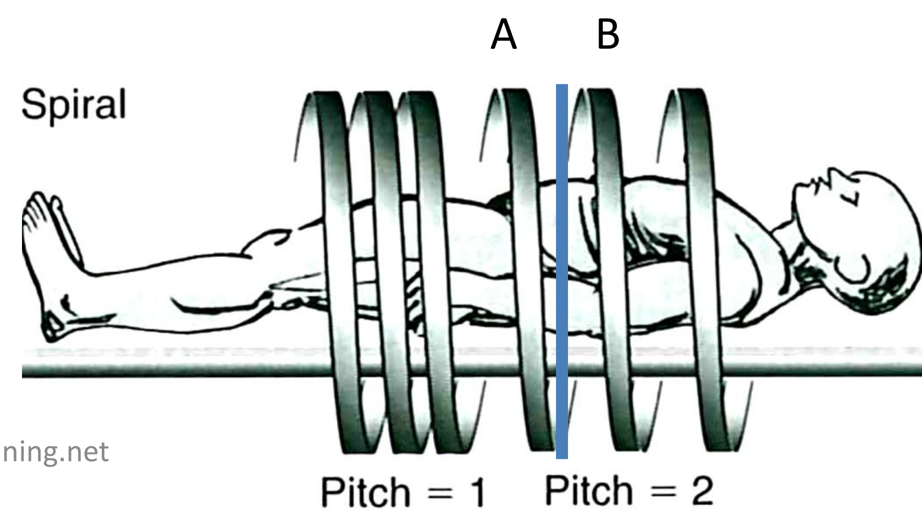
- **VOL. COVERAGE =**

- **pitch x slice thickness (collimation) x number of rotations**

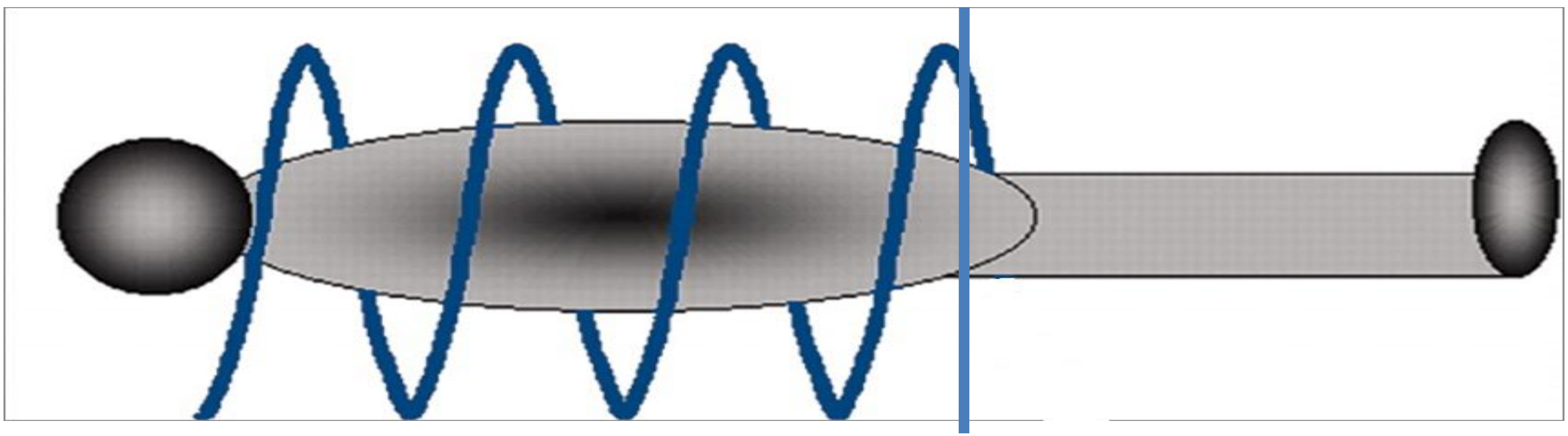
Data interpolation in helical scanning

- Definition: Estimation of a value using known values on either side
- In helical CT: volumetric data is collected , then reconstructed into images
- To reconstruct the slice drawn in the image: there must be interpolation of data collected at position A & B (relative weight given to each is dependant on relative distance from plane of reconstruction)





Problem: if the reconstructed cut is at the start or the end of the scan → no data are available for interpolation
Solution: extra rotation at the beginning and the end of the scan for the purpose of interpolation



Advantages of helical scanning

1- permit faster rotation time (continuous rotation) → increase of the scan speed

- Less slice mis-registration (single breath hold)
- Less Patient motion artifacts
- Reduced dose of contrast medium

2- permit continuous data acquisition while the table moves → data collected is a complete volume not a single slice (No gaps in data acquisition) = volumetric imaging

- slice can be reconstructed for any axial position

Notes:

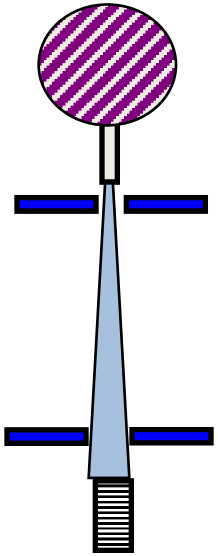
1-width of reconstructed section could not be less than the detector width (collimated beam width = slice thickness)

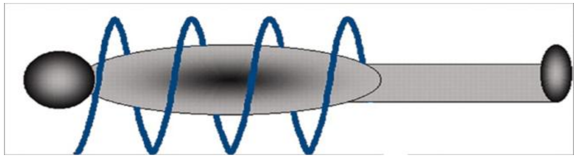
2- Effects of increasing the pitch above 1:

- Less exposure time
- Decreased patient's dose
- needs greater interpolation → loss of resolution

3- For most of clinical applications , pitch is not increased beyond 1.5

4- helical CT tubes must have high heat rating (longer continuous exposure times than conventional scanners)

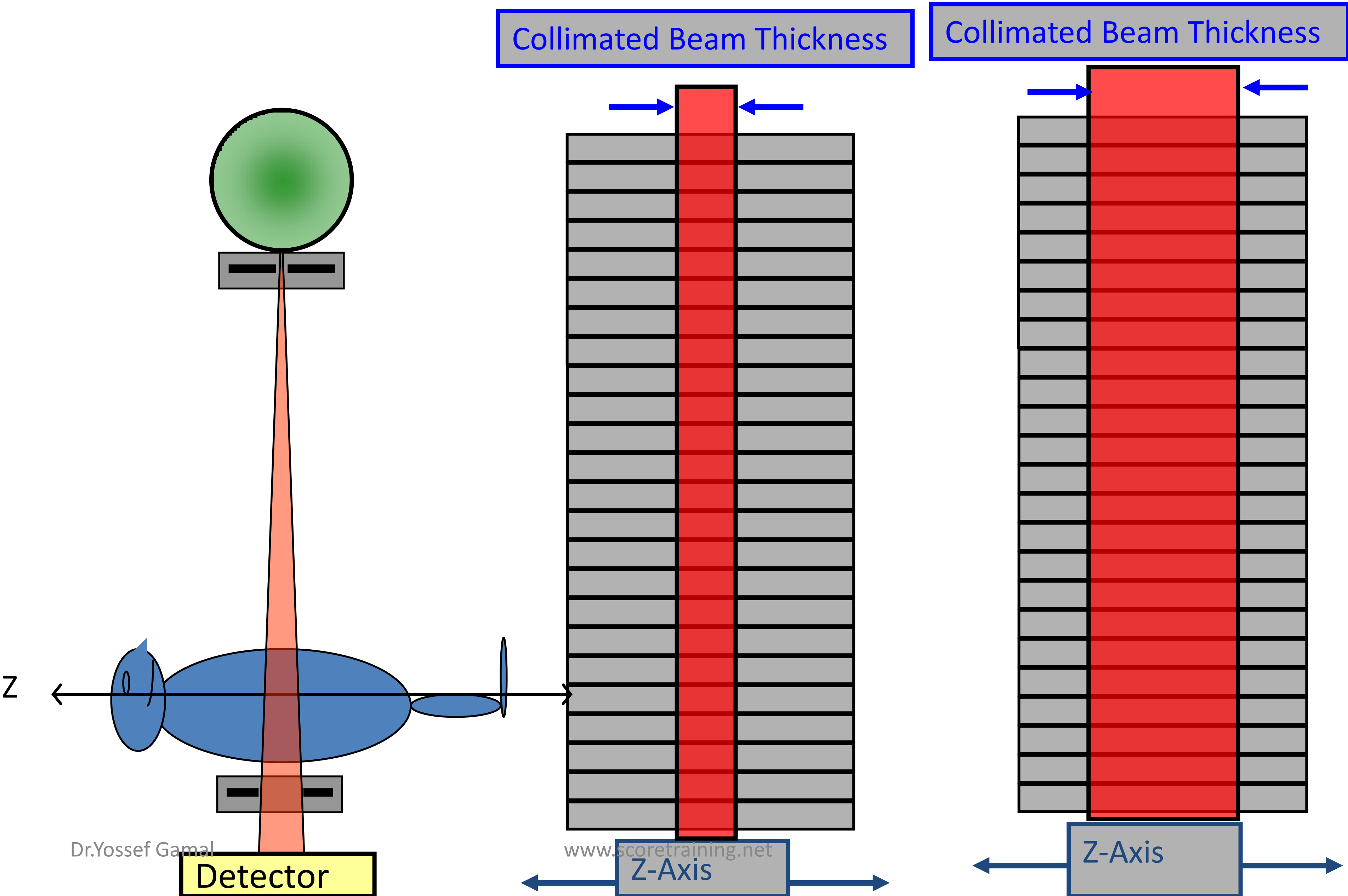




Multi-Slice CT

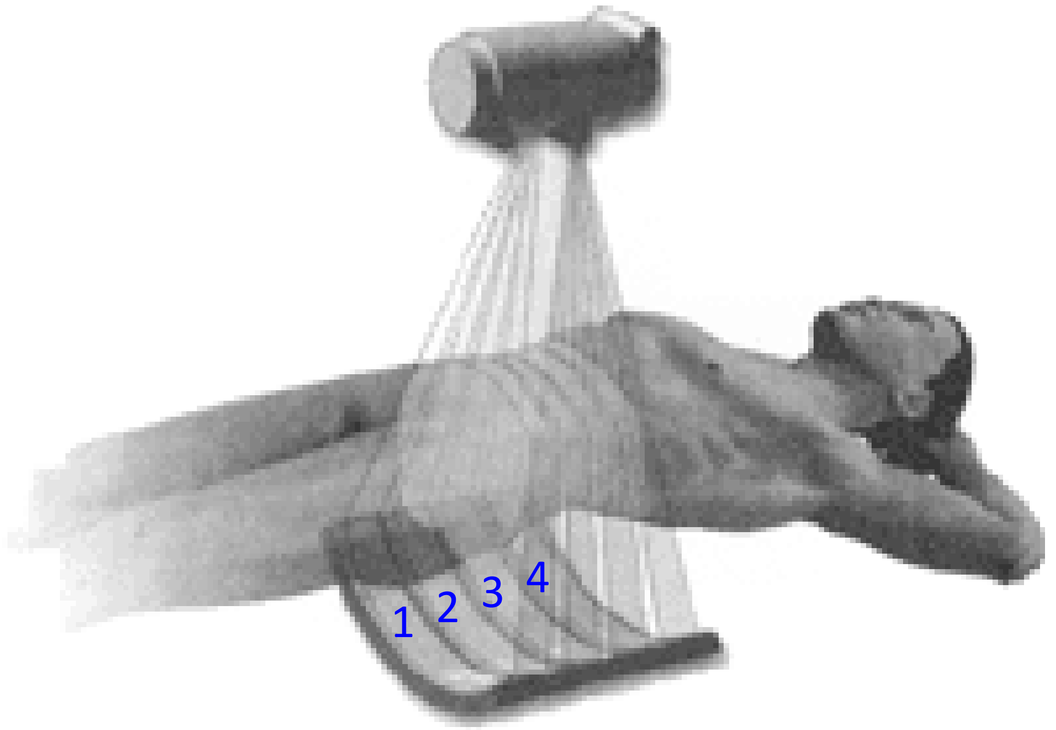
In Single Slice CT:

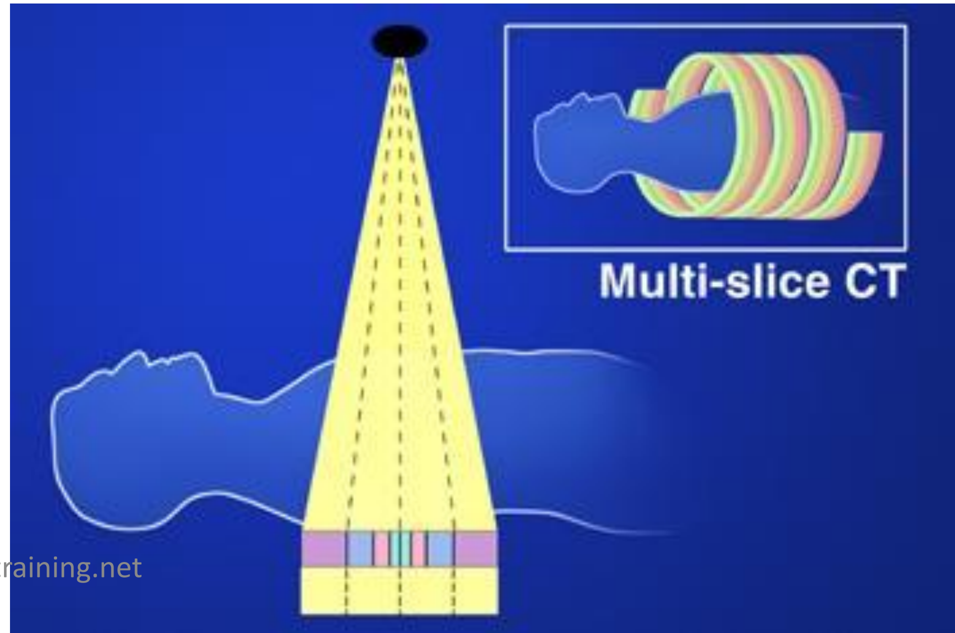
1- There is Single row of detectors in z-direction
2- Slice thickness is Determined by Collimation



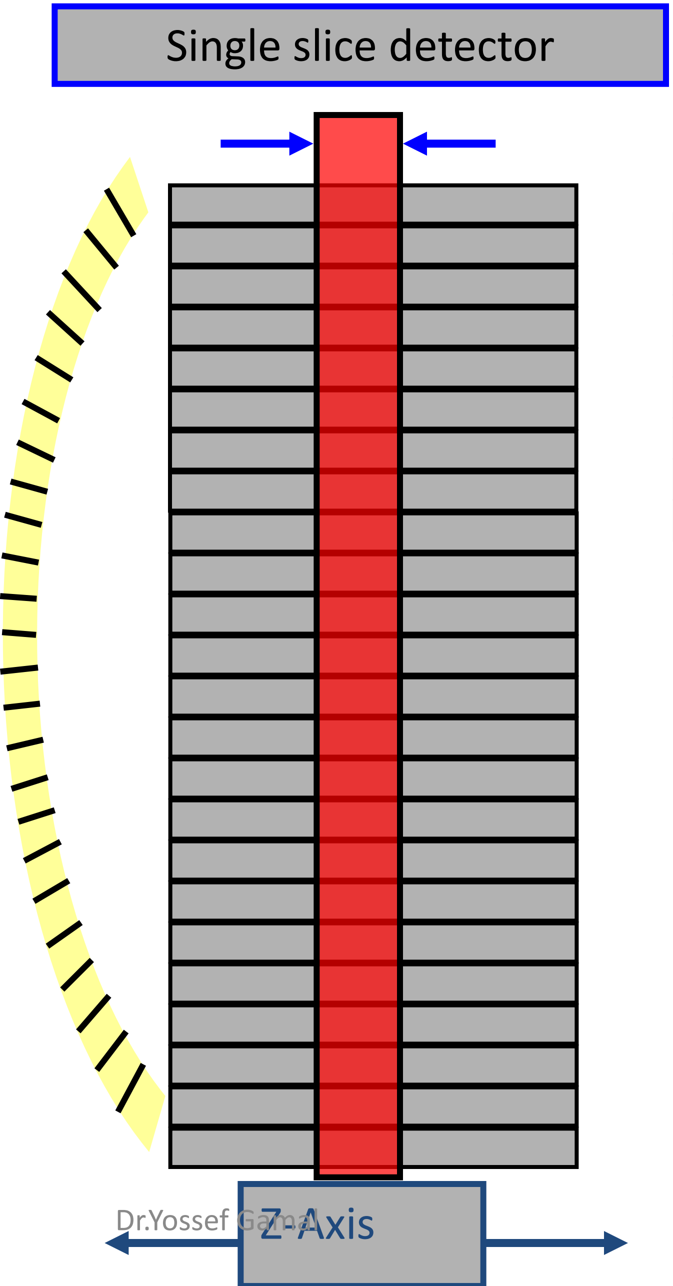
Multislice CT

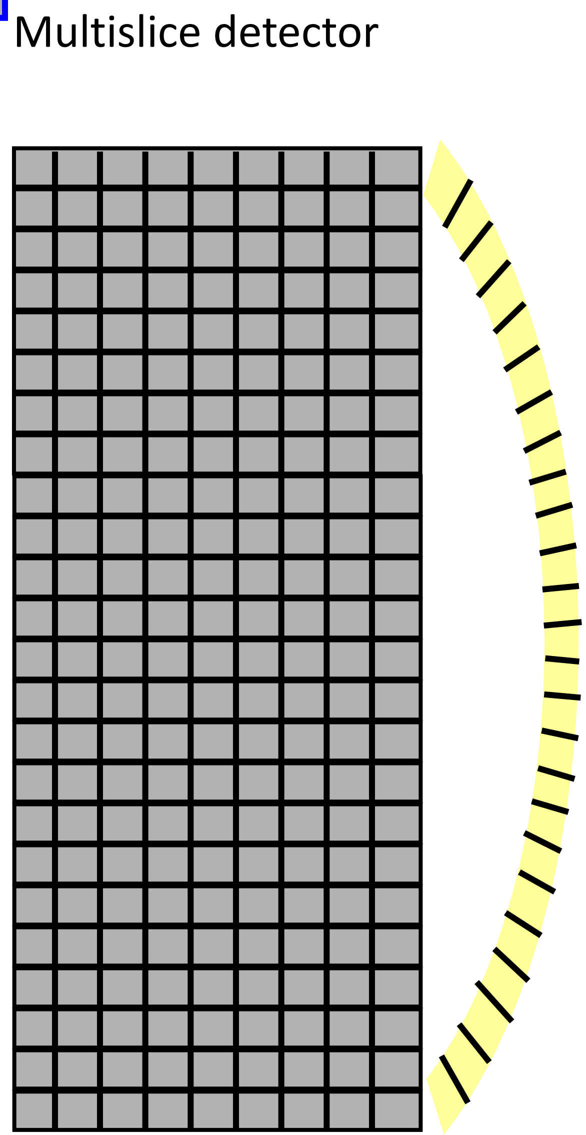
- Multiple rows of detectors arranged in Z direction
- Simultaneous acquisition of multiple slices

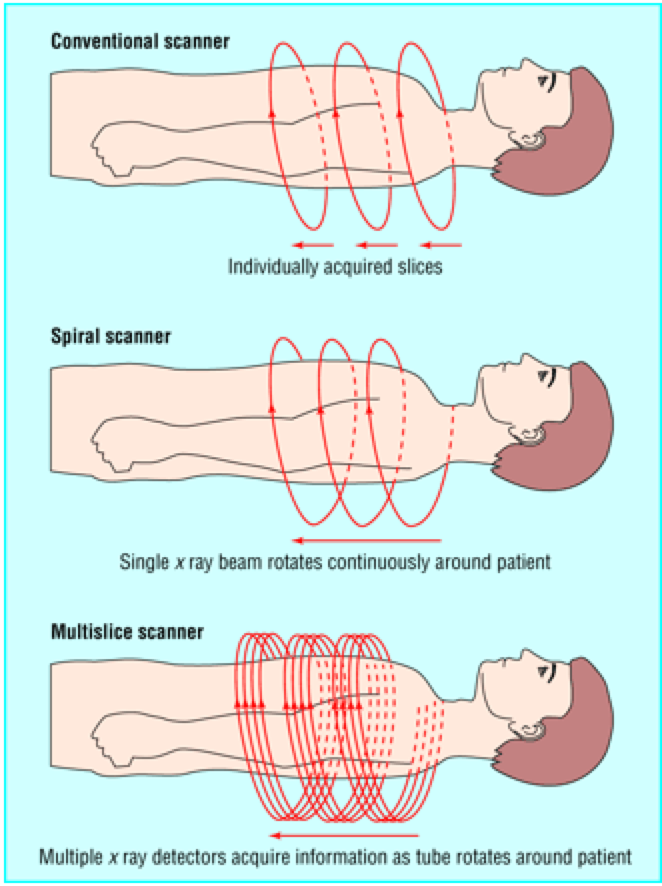




Single Slice vs. Multislice Detector







Dr.Yossef Gamal

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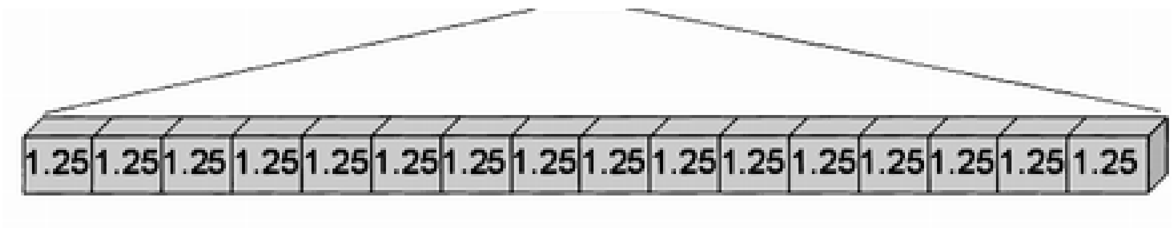
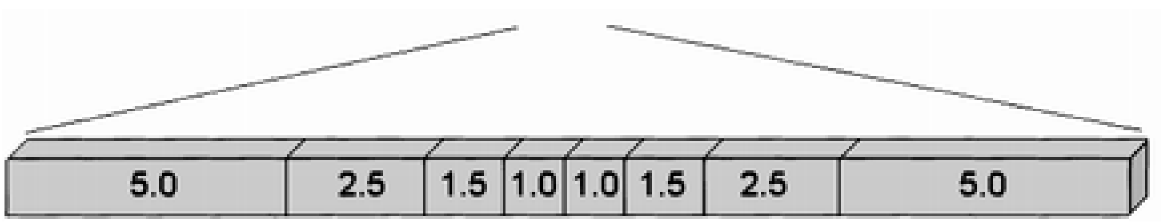
- multislice CT scanners detector array designs is divided into two groups:

1- matrix detectors array: detector elements are of equal width along the z axis



2- adaptive detectors array: detector elements are of unequal width

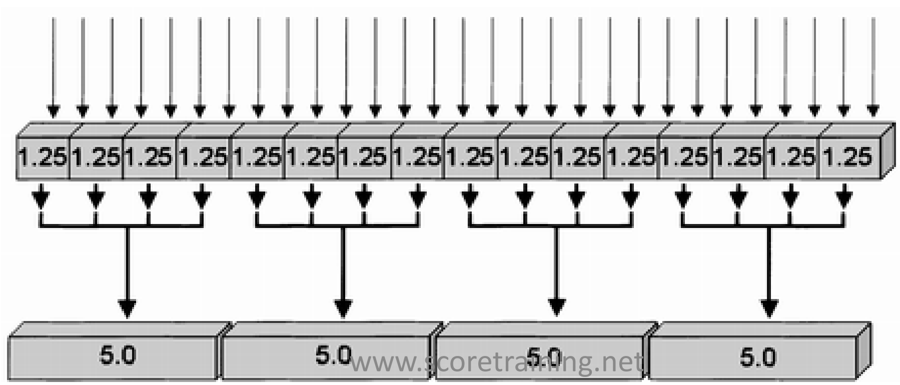


Data Channels of multislice scanners

- The number of sections that are acquired depends on the number of z-axis data channels on the CT scanner.

i.e. number of sections that are scanned per rotation ≠ the number of detector rows on a scanner

i.e. a four-channel multi–detector row CT scanner has four z-axis data channels and enables the acquisition of up to four sections per rotation, even though many more detector rows are present.



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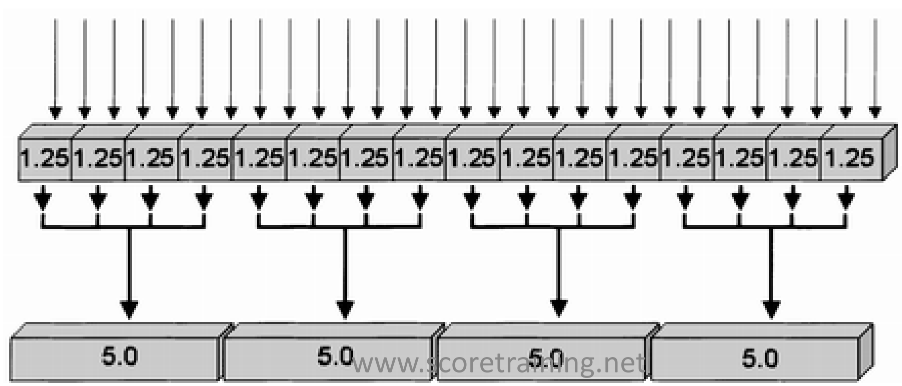
Effective detector row thickness

- The sum of the widths of the contributing detector rows for each channel

= scanned slice thickness

- The reconstructed section thickness cannot be smaller than the effective detector row thickness



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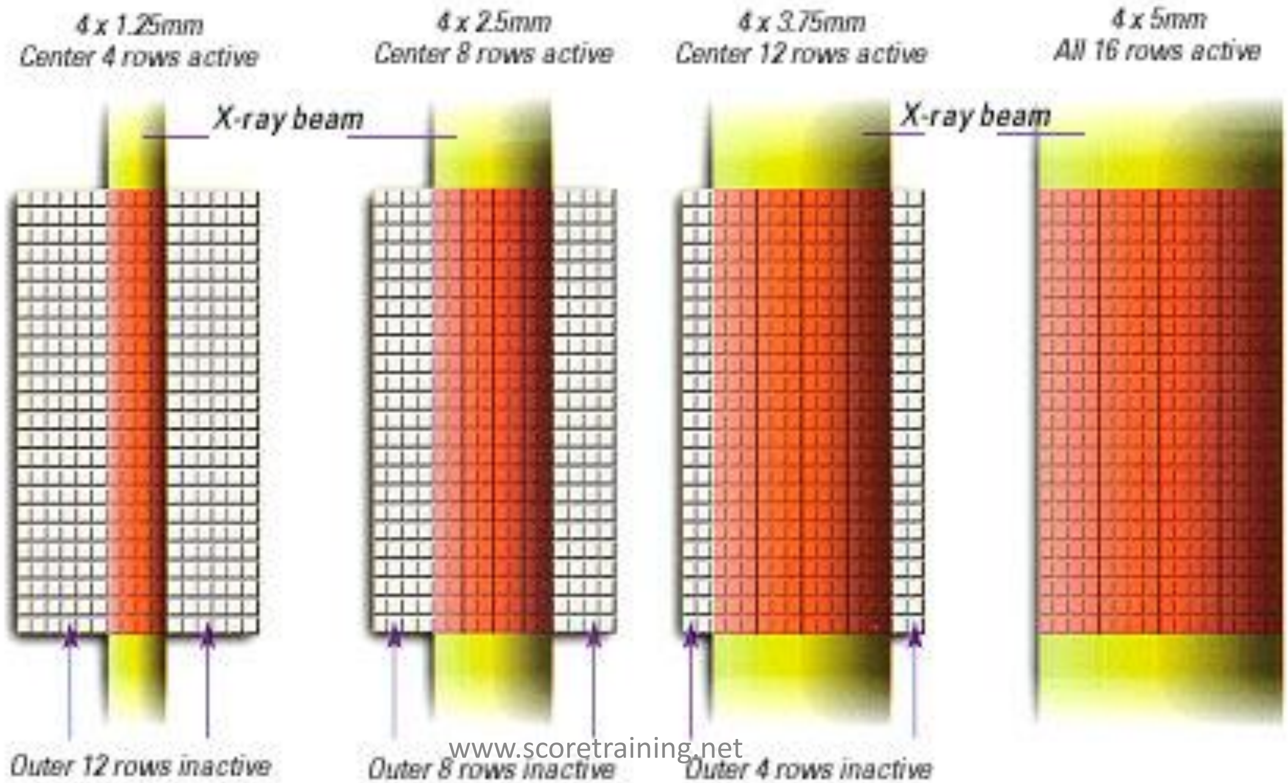
Detector Configuration:

- number of z-axis data channels X effective detector row thickness of each data channel.

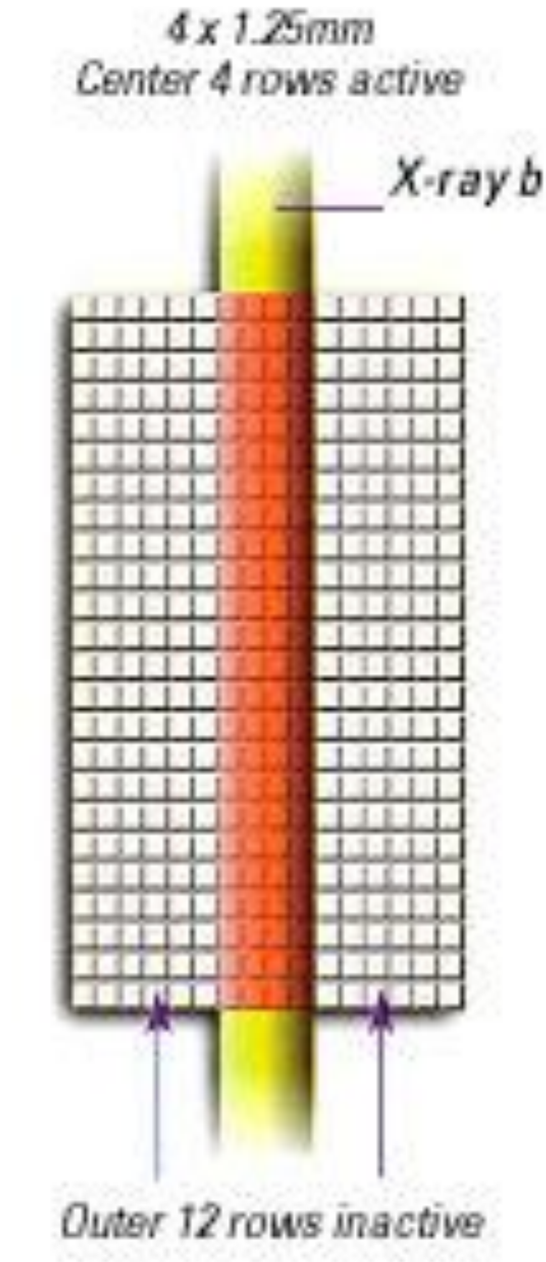
e.g. a detector configuration of 16 x 1.25 means:

16–data channel acquisition performed with effective detector row thickness of 1.25 mm.



Beam Collimation:

- The collimated beam length= number of data channels X effective detector row thickness.
- for a 16 x 0.5-mm detector configuration→ beam collimation = 8 mm
- Practically : the collimated length will be somewhat greater to ensure that the outer 2 rows are uniformly covered by the x-ray beam

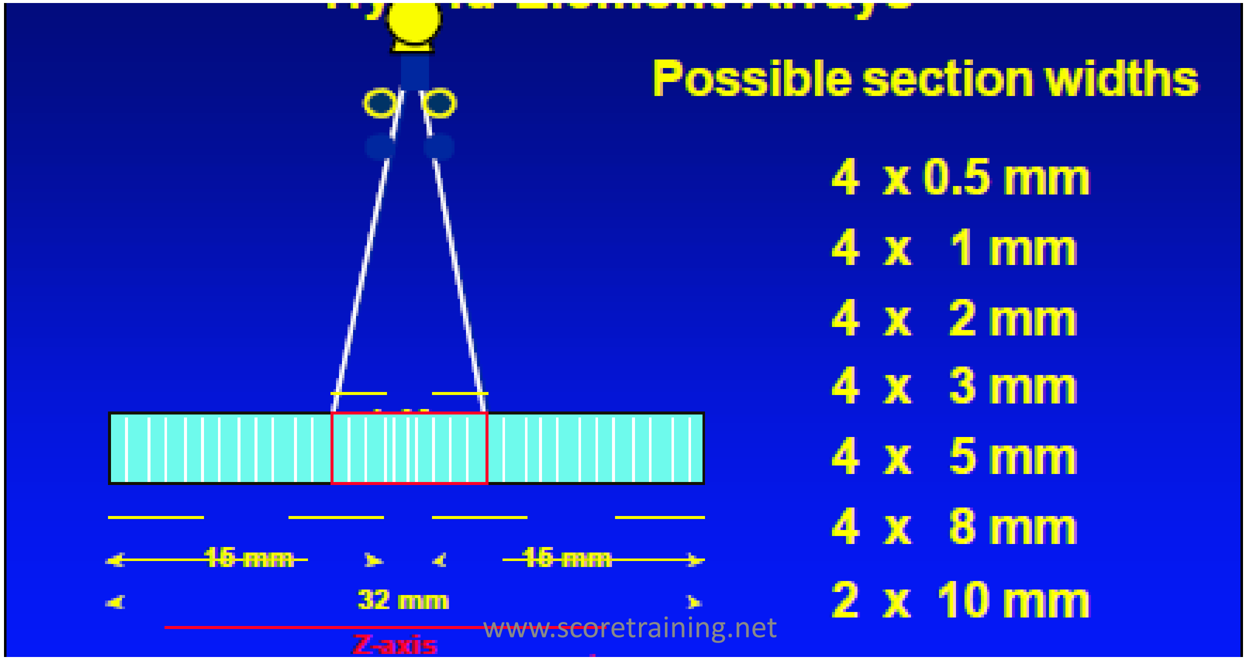


In this example collimated beam =………

Example of matrix detector array

Multislice CT system with:

- 20 detectors
- Every detector is 0.5 mm thick
- 4 channels



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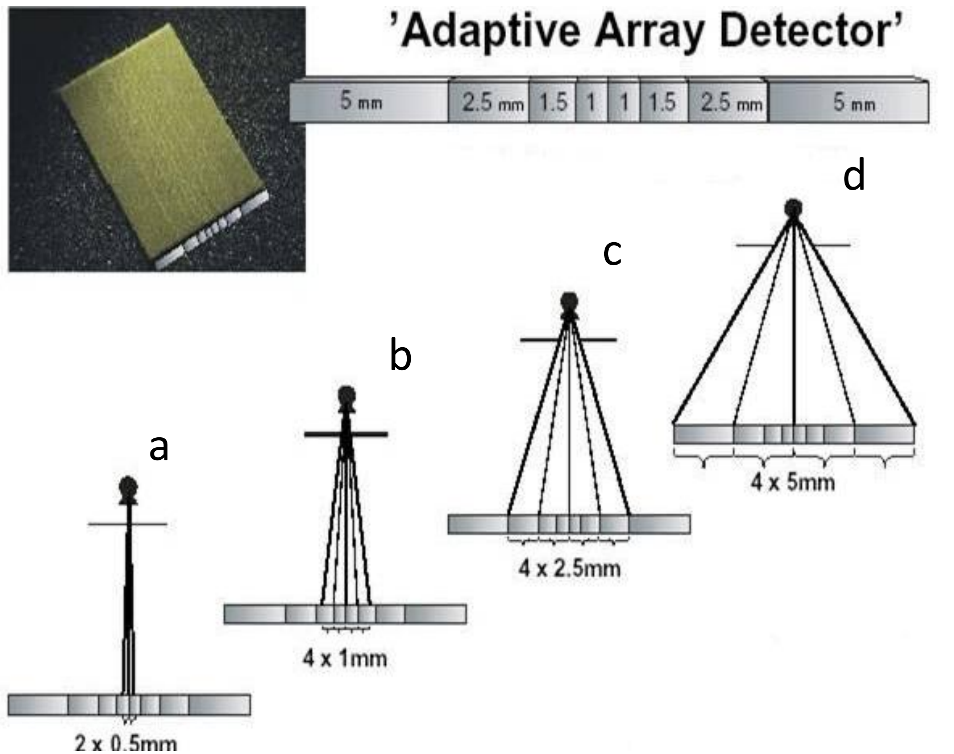
Example of adaptive array detectors

Multislice CT system with:

- 8 detectors
- Central detectors length is smaller than the peripheral detectors
- 4 channels

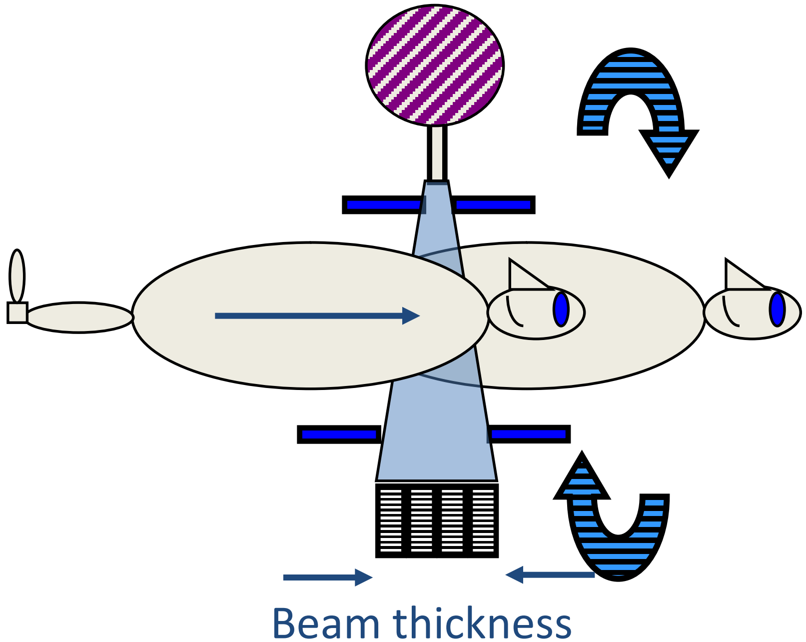


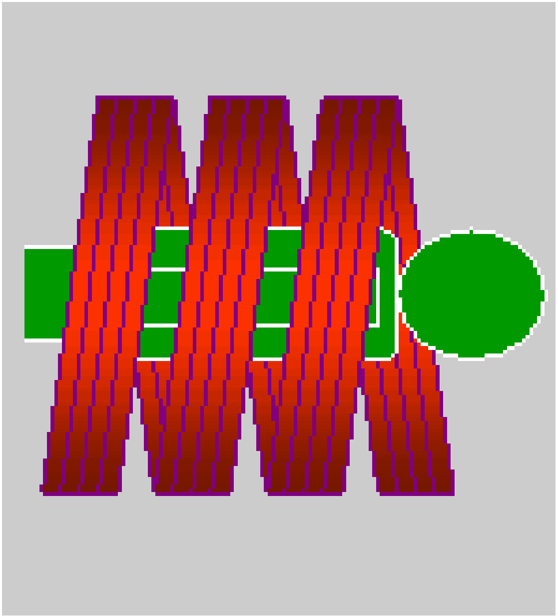
In option a : collimated length = 1 mm
In option b: collimated length = 4 mm
In option c : collimated length = 6 mm
In option d: collimated length =20 mm

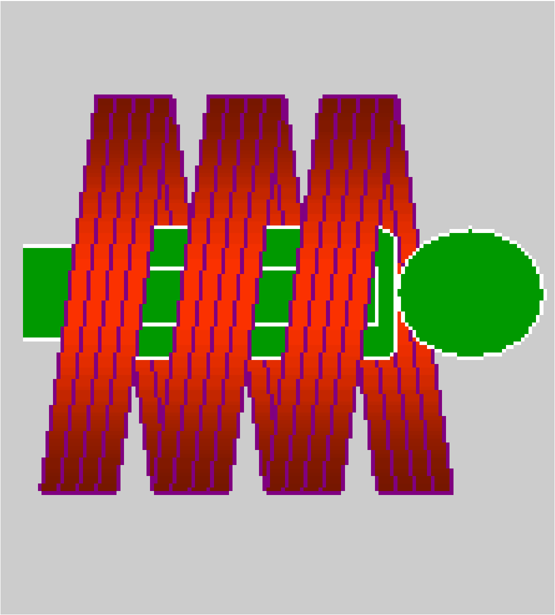
Multislice Beam Pitch

$$\text{Beam Pitch} = \frac{\text{table motion during one rotation}}{\text{Beam thickness}}$$

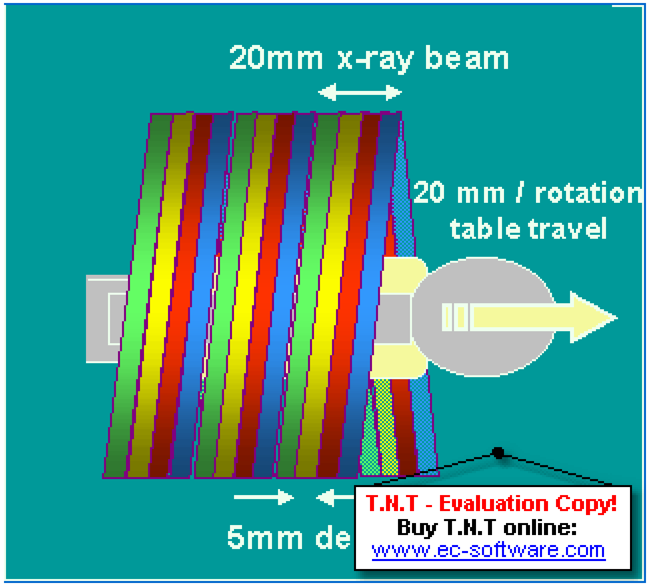
Note that slice thickness is replaced by beam thickness (collimated thickness)







Beam Pitch > 1
Gap in data colliction



Beam Pitch = 1
No gap

Single Slice Pitch Definition

$$\text{Slice Pitch} = \frac{\text{table motion during one rotation}}{\text{slice thickness}}$$

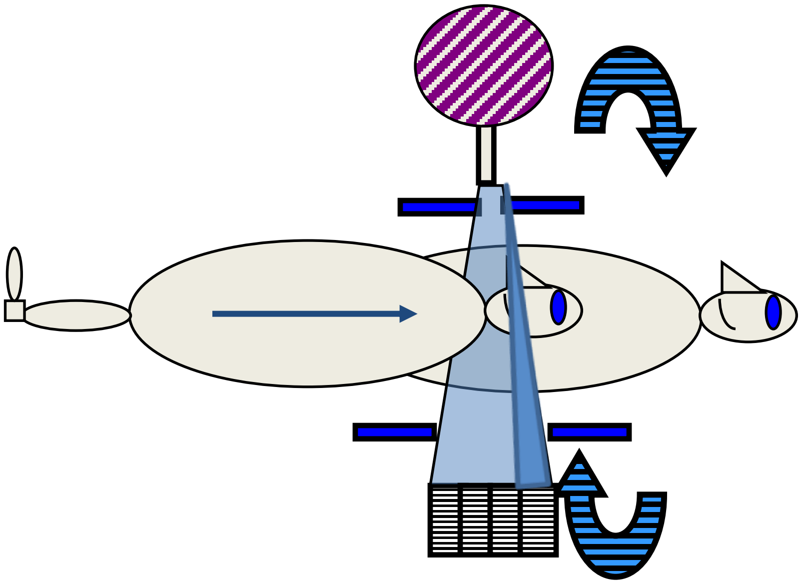
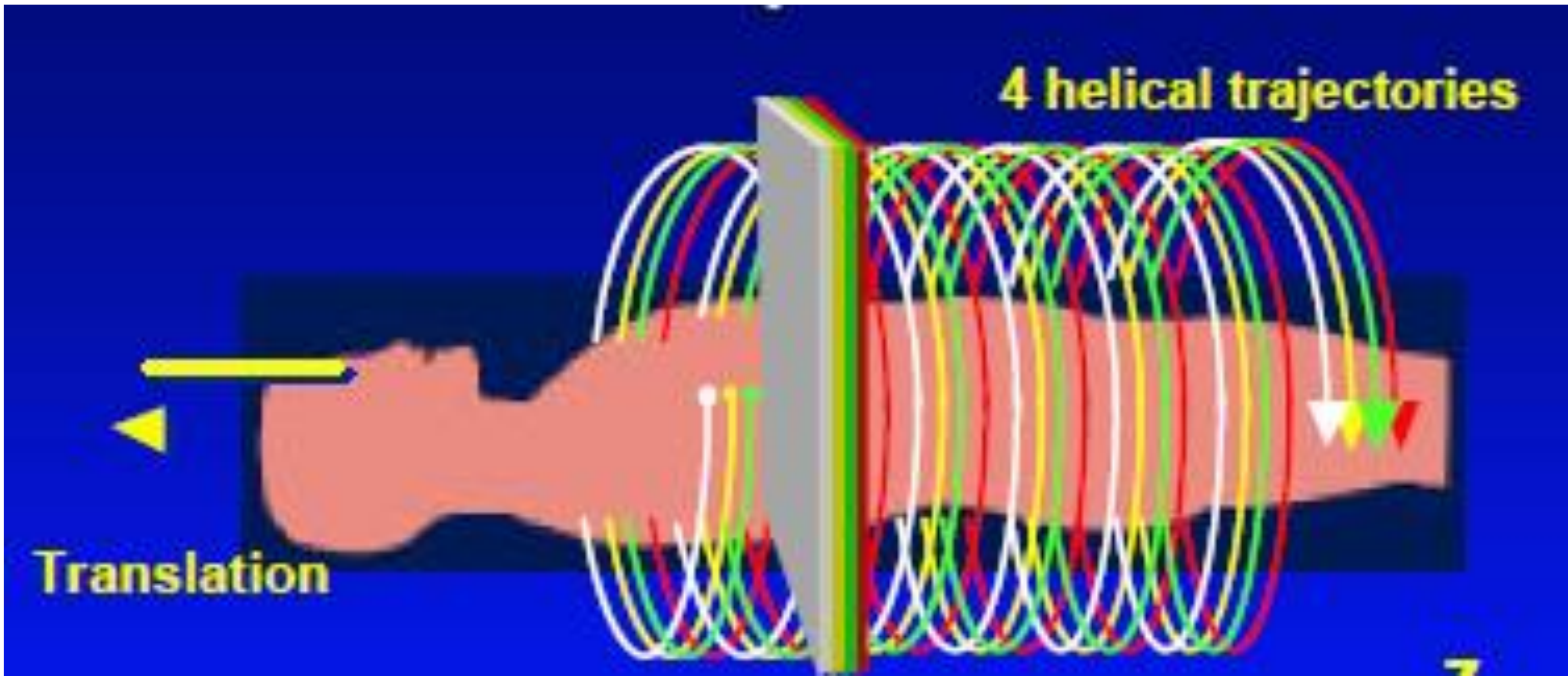


Table Travel Speed

= beam collimation x beam pitch x number of gantry rotations per second.



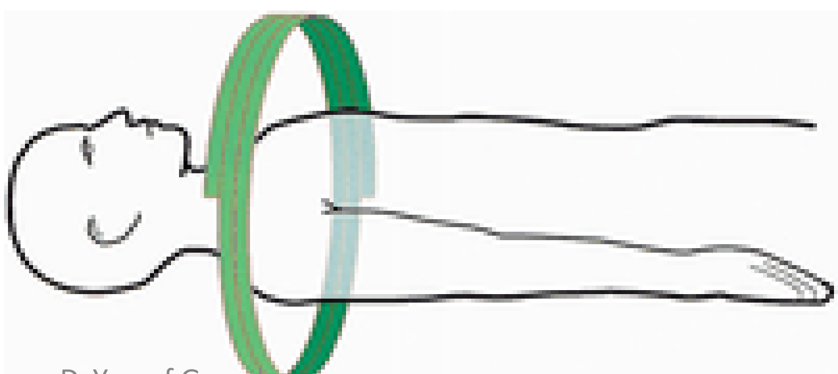
- **Advantages of multislice scanning**:

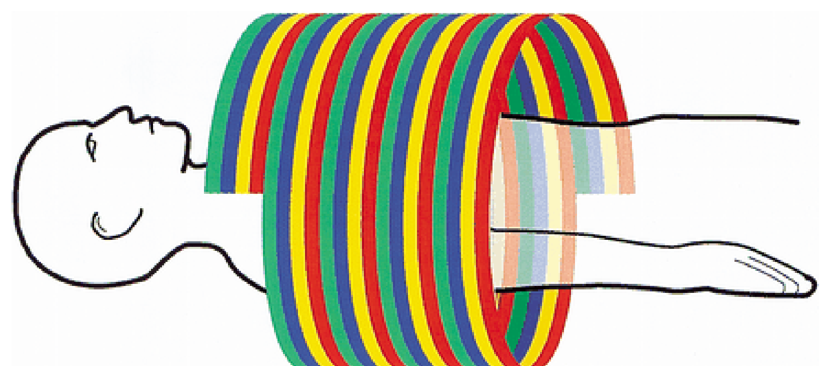
1) Faster scanning:

- Because multisection CT scanners can generate multiple sections per revolution, and because of faster gantry rotation speed

→ fewer motion artifacts, less contrast medium used... etc.

2) Longer anatomic coverage: same causes

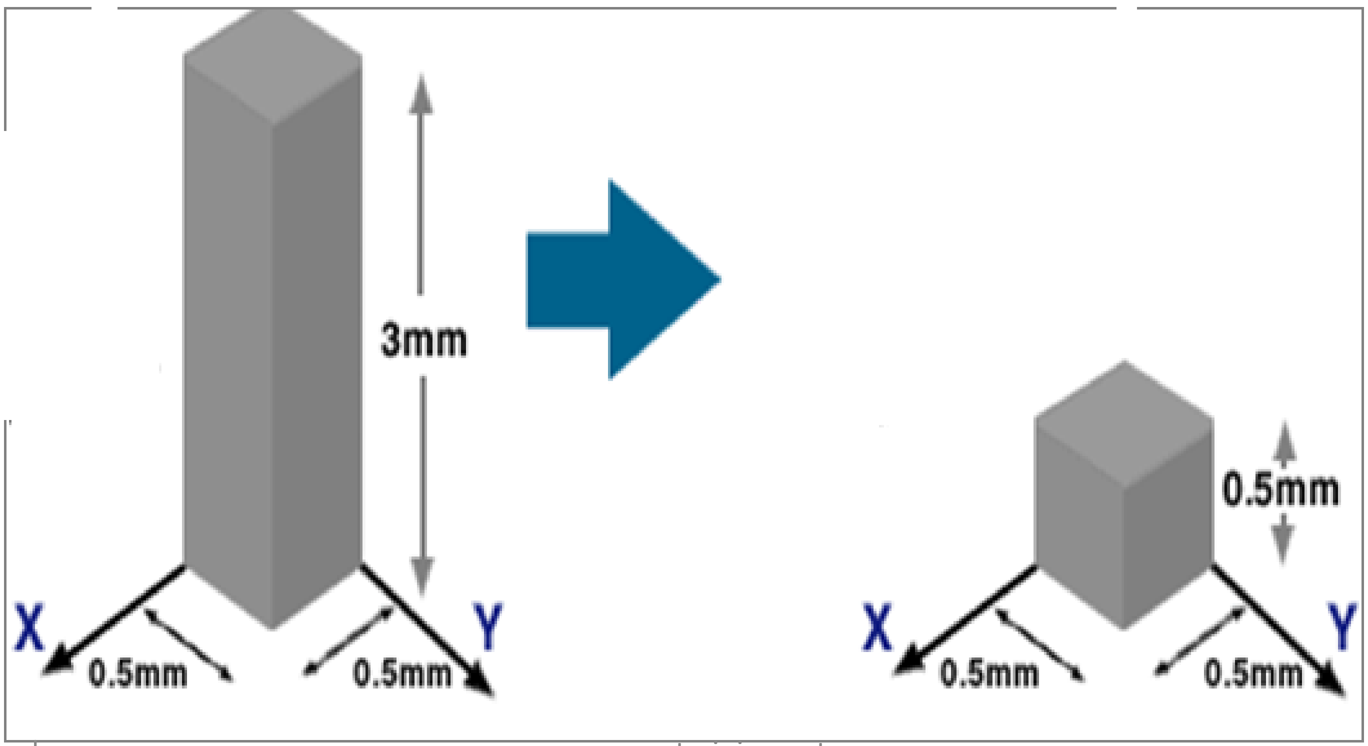




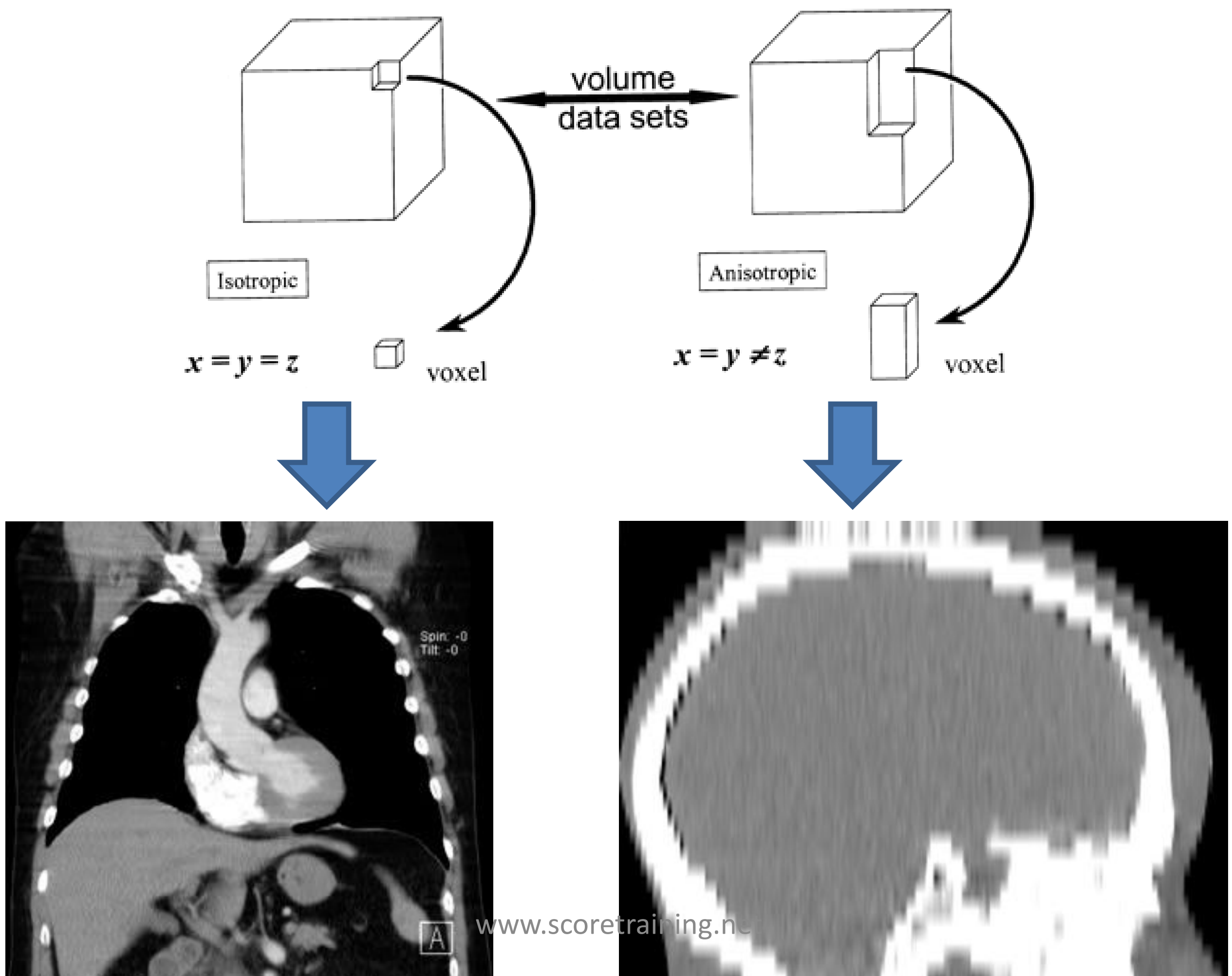
3) Isotropic imaging with improved spatial resolution:

Isotropy: dimensions of each voxel is equal in all of the three planes

Anisotropy: z dimension of the voxel › x & y dimensions



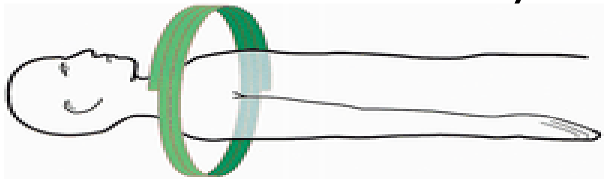
Isotropy allows reformatting the image in any plane with spatial resolution identical to the original scanning plane



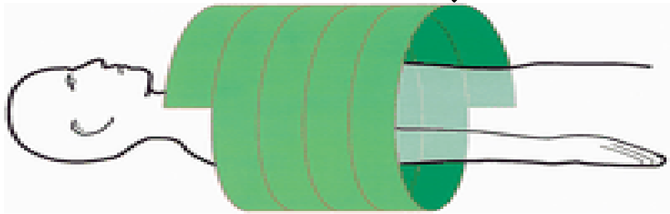
- Voxel size in transaxial plane (x & y):
  - same for single and multi-slice imaging
  - i.e. depends on matrix size and the field of view
  - typically in the region of 1 mm
- Voxel size in z plane:

1- for helical (single slice) scanners:

A- if isotropic imaging is required (beam width = 1 mm) → small volume covered (tube permit only total scan time of about 90 s.)

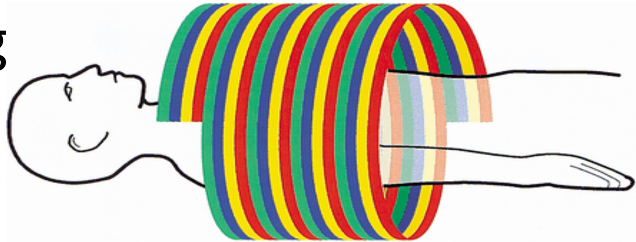


B- if large volume is imaged → slice thickness must increase → anisotropic imaging



2- for multislice CT:

slice thickness can be easily decreased to 1mm combined with good volume coverage → isotropic imaging



.

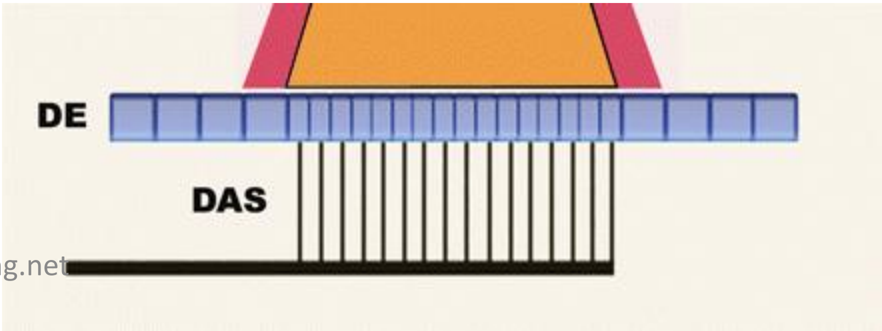
- Notes:

1-a gap between detectors is present to prevent light crossover between detectors and to reduce effects of scatter produced in the detector

2- in multislice CT no post-patient collimation is used

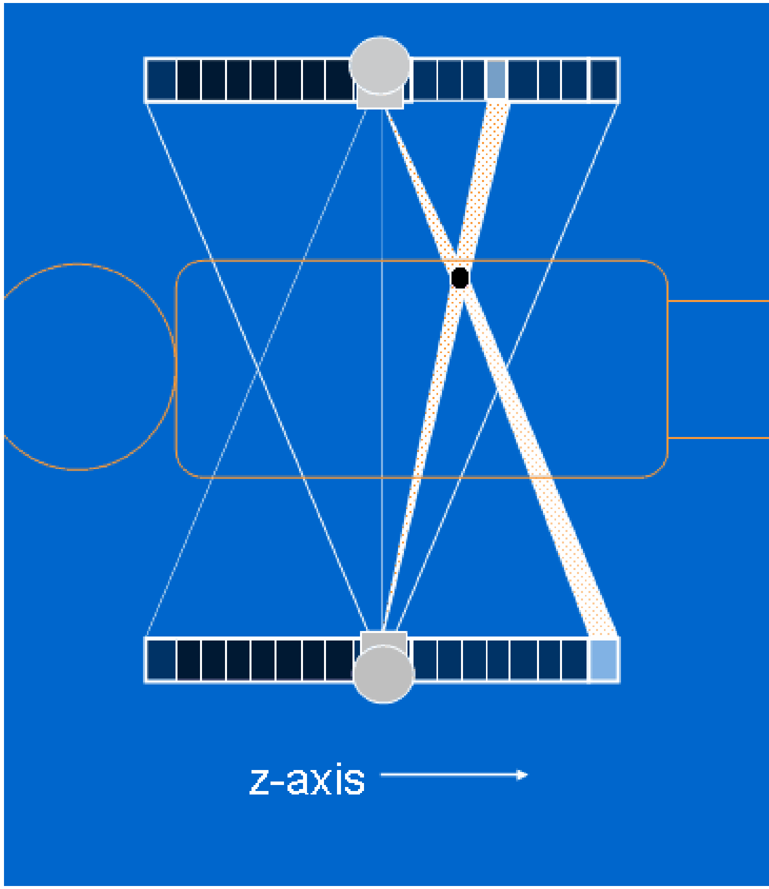
→a portion of the beam (rose bands) must extends beyond the active detector elements.

This area of overextension, called the penumbra, is necessary to ensure exposure of the most peripheral of the active detector elements



Cone beam effect in multislice scanners

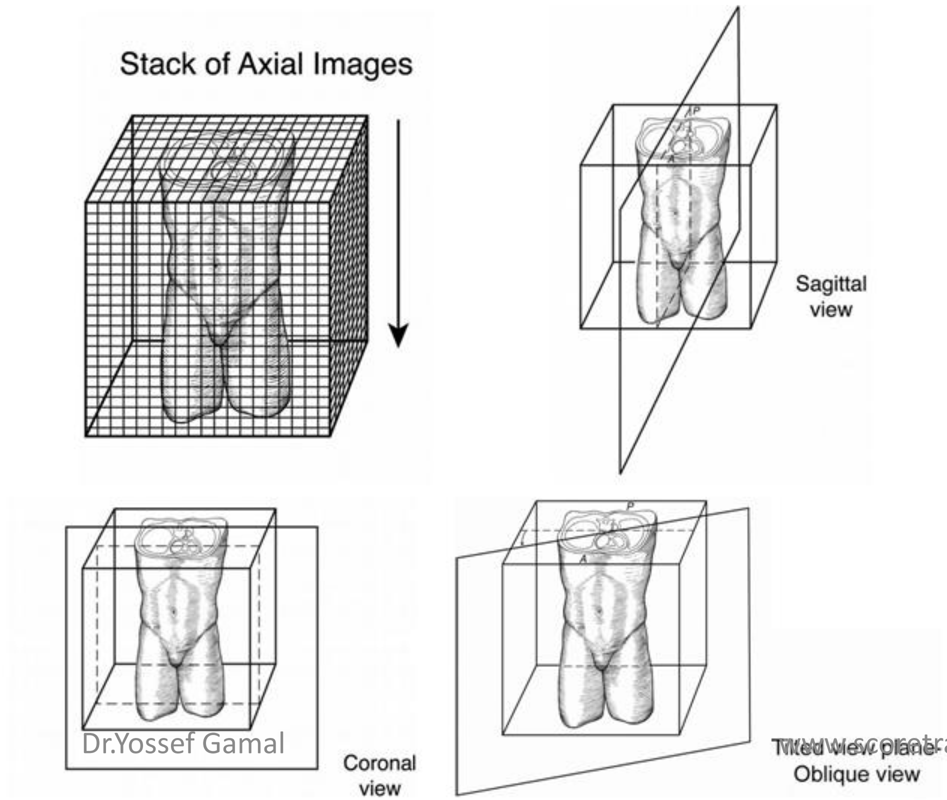
- Off axis objects are seen by different detector rows for different tube angles
→misregestiration of data
→ image artifacts
- Caused by beam divergence
- Effect is getting worse with :
  - Increased number of slices/rotation i.e. 16 slice scanners › 4 slice scanners
  - Increasing total detector length
  - Multislice scanners (The problem is minimal for helical scanners)



CT image post-processing

1- Multiplanar reformation (MPR):

- It is the process of using axial data to create non-axial two-dimensional images(formed from only 1 voxel thickness)
- It may be sagittal , coronal or oblique



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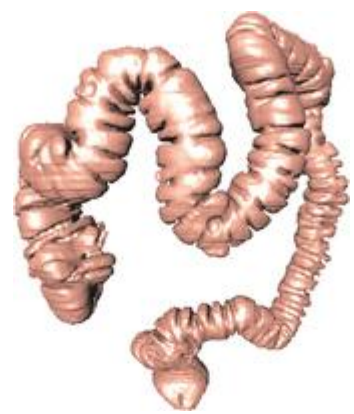
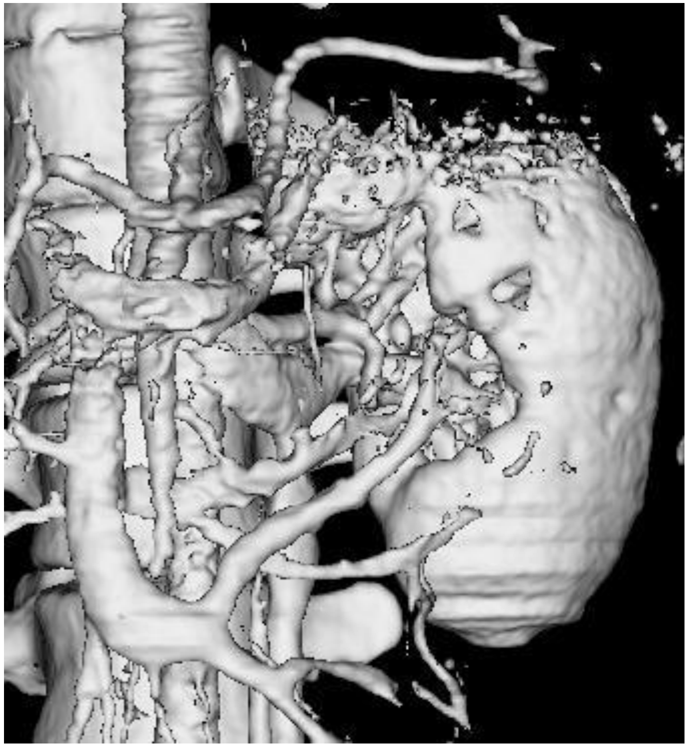
2) surface rendered image:

- Surface structures are displayed by selecting suitable CT numbers with upper and lower thresholds, the remainder of the data is discarded.

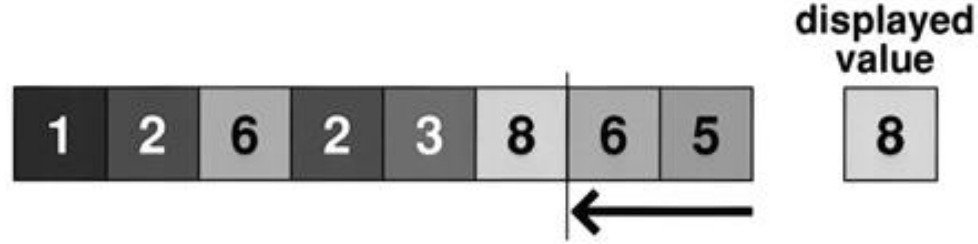


3) shaded surface display:

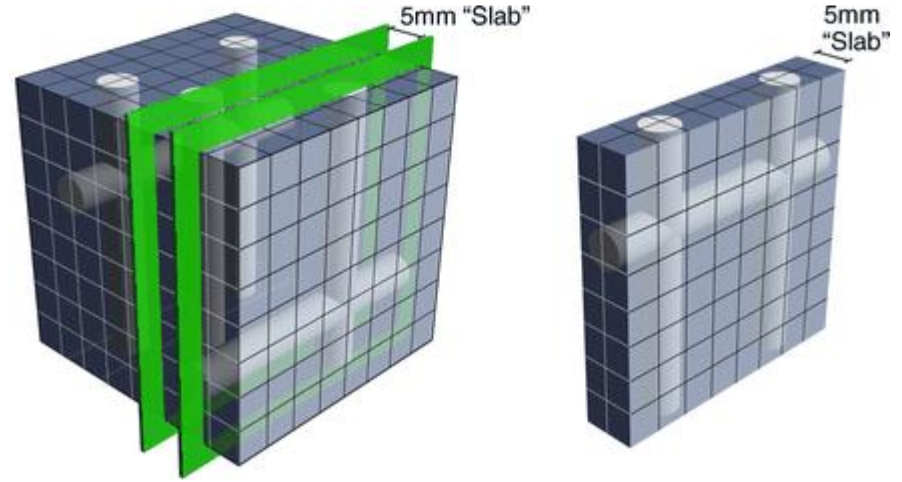
- As previous but with shading (considering effect of virtual light source on the structure)

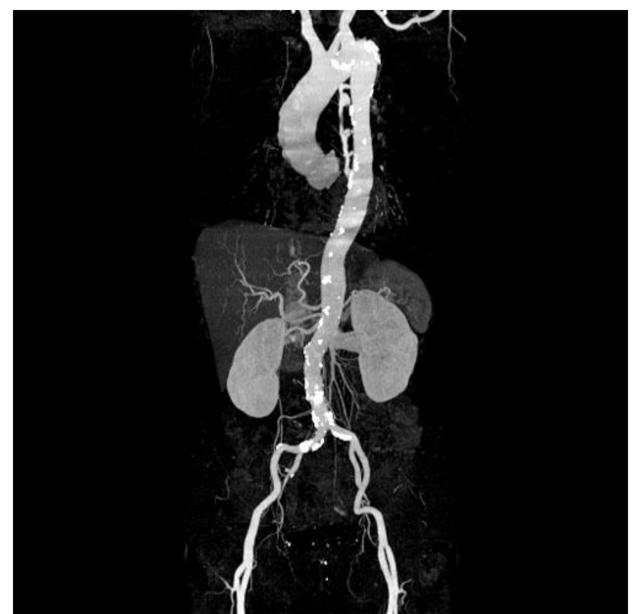


4) maximum intensity projection:



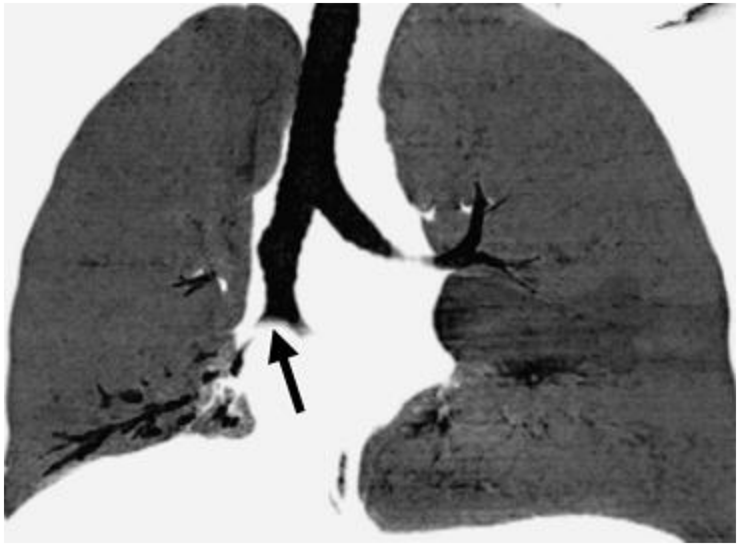
- MIP images are achieved by displaying only the highest attenuation value from the data encountered by a ray cast through an object to the viewer's eye (the background is subtracted)
- Used in CT angiography





5- minimum intensity projection:

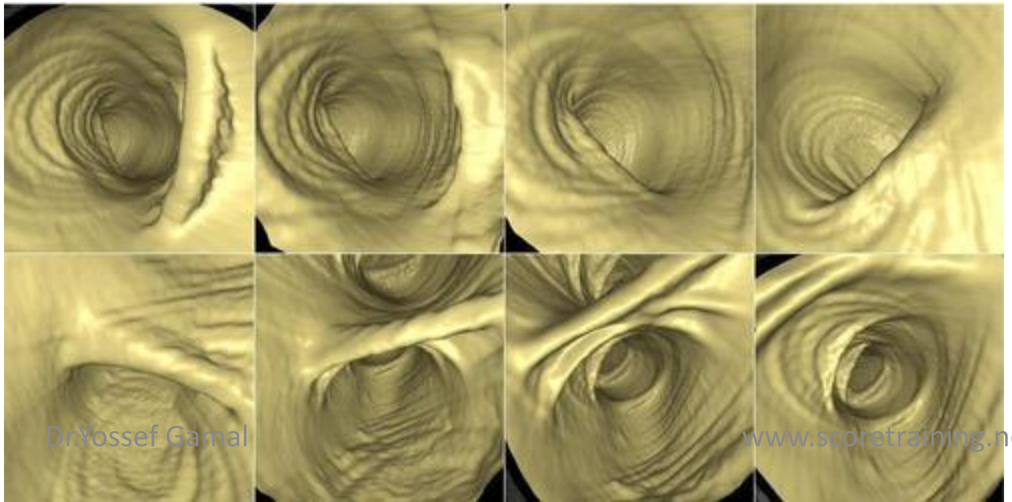
- As above , but the minimum CT numbers are displayed
- Used to visualize tracheobroncheal tree



6- 3D volume rendering:

- CT numbers that make up the image are displayed with varying colors, and with varying opacity levels

N.B: virtual endoscopy is a type of 3D volume rendering

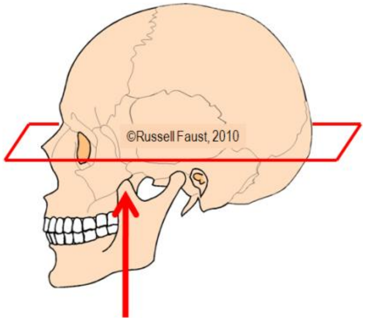


CT image quality

CT spatial resolution

- Maximum spatial resolution that can be achieved in CT is 20 lp/cm (less than conventional radiology)

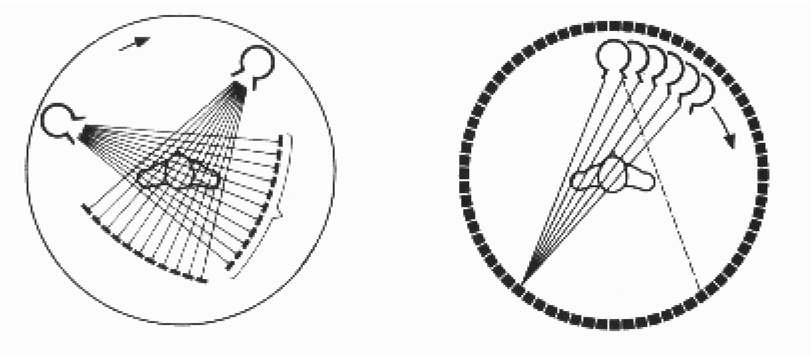
Factors affecting CT Spatial Resolution



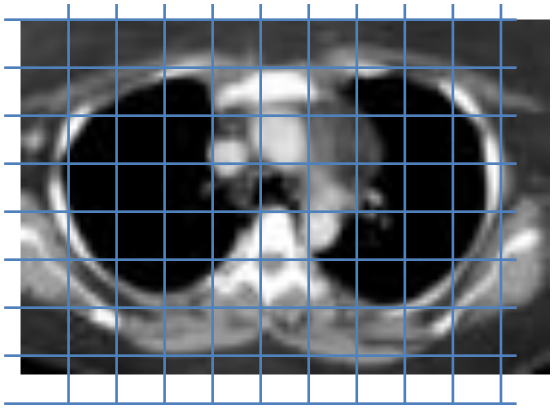
1) Resolution in the transaxial plane

<u>A- Non-geometric factors:</u>

- # of projections/slice
- pixel size (= FOV/matrix size)
  - i.e. sampling frequency
  - 512 X 512 pixels standard



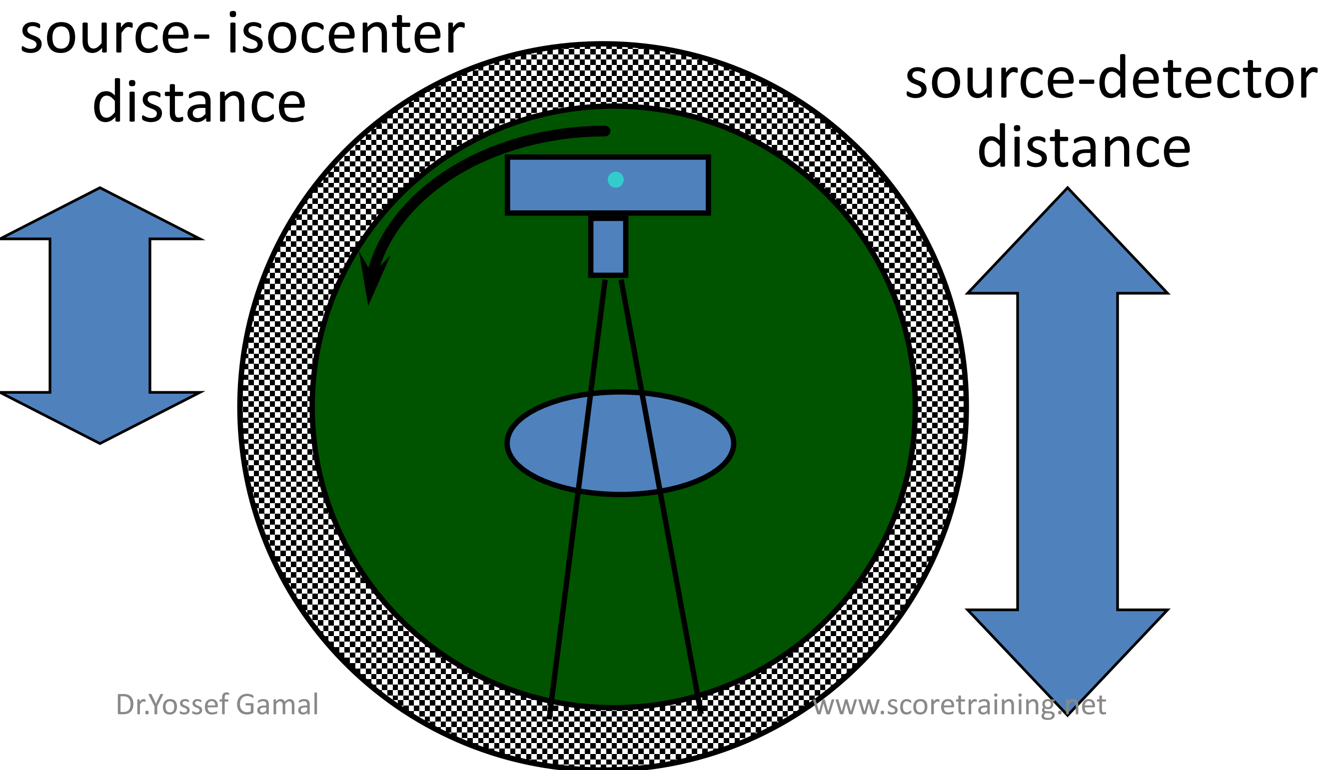
The previous factors are Not intrinsic limitation i.e. can be modified

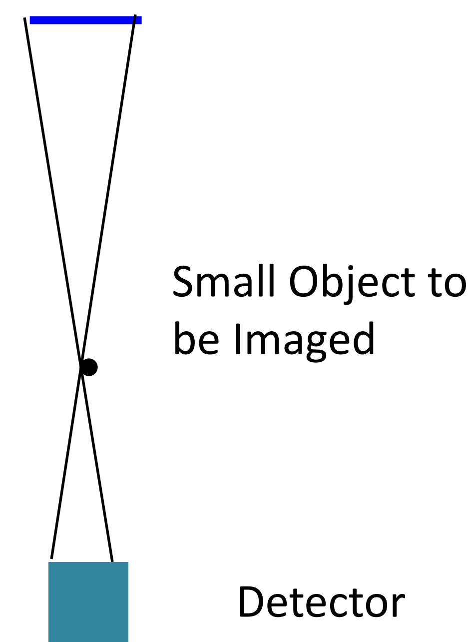


B-geometric factors:

- Focal spot size
- Source to detector distance
- Source to isocenter distance
- Size of sensitive area of the Detector (object must be larger than this to be resolved)

The previous factors are intrinsic limitation   i.e. can not be modified



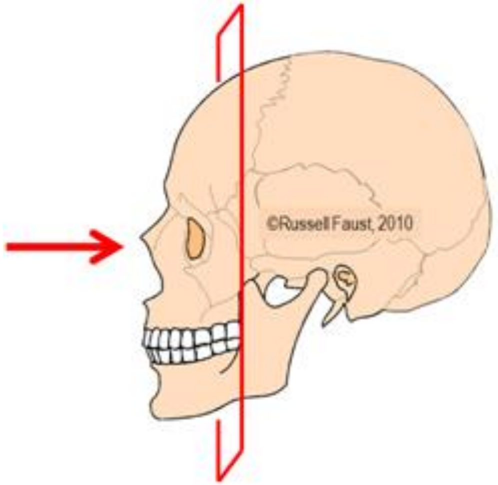


C-Reconstruction Algorithm

- Back projecting process blurs image → decrease the spatial resolution
- Special algorithms: Can increase or decrease edge definition
  - edge enhancement
  - Low pass filter

N.B: the previous factors are not significantly different for axial and helical scanning

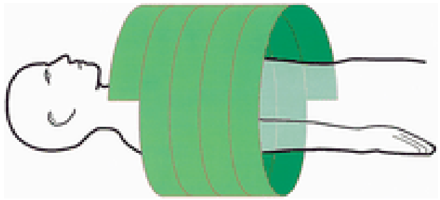
2) Factors affecting Resolution in the Z plane (resolution of the reconstructed images):

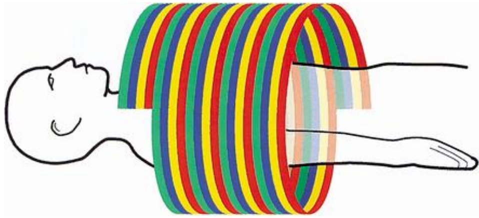


- slice thickness

=……….. for single slice

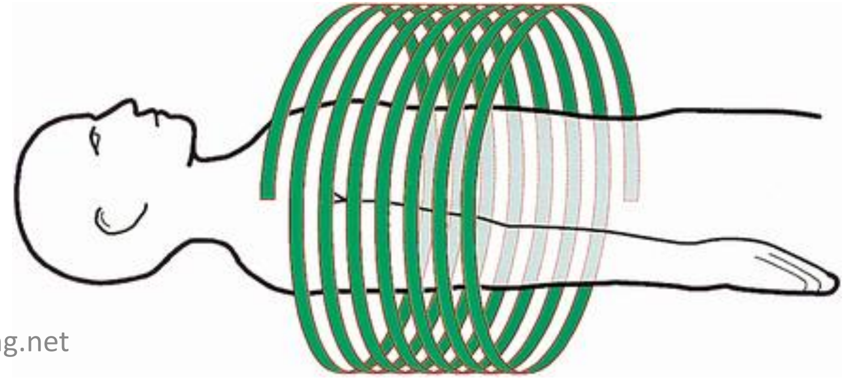
=………… for multi-slice





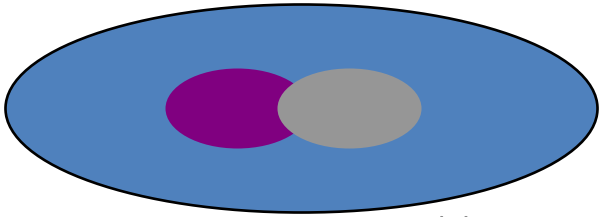
i.e. better for multi-slice scanners

2- pitch: ↑pitch →↓ spatial resolution



Contrast Resolution

- Ability of an imaging system to demonstrate small changes in tissue contrast
- The difference in contrast necessary to resolve 2 areas in image as separate structures
- CT Contrast Resolution Depends on Noise and reconstruction algorithms



CT Contrast Resolution

- Significantly better than radiography
- CT can demonstrate very small differences in density and atomic #

Contrast Resolution

Radiography	CT
10%	<1%

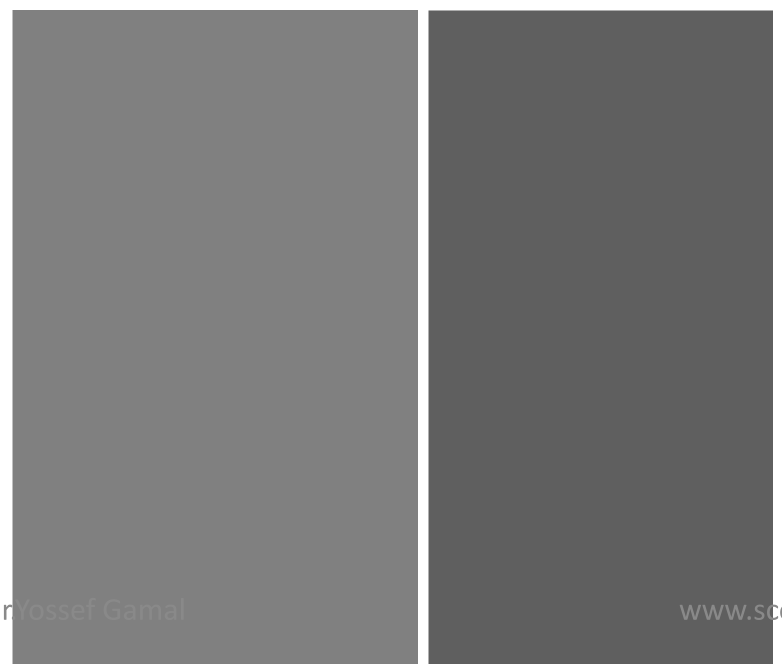
CT image noise

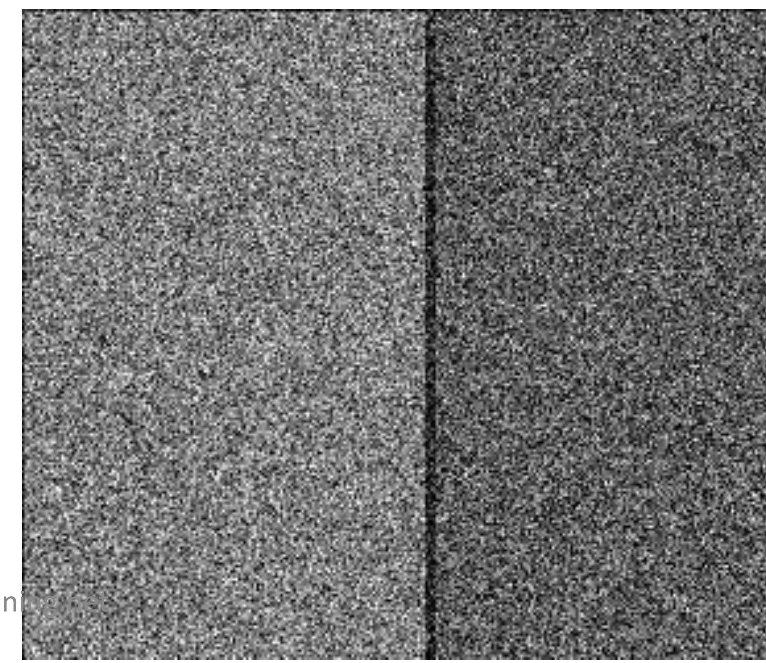
- High noise will cause decrease of contrast resolution
- Sources of noise in CT:

1-Quantum noise:

2- electronic noise

3- structural noise





Quantum noise:

Depends on the number of detected photons

SNR α √number of photons

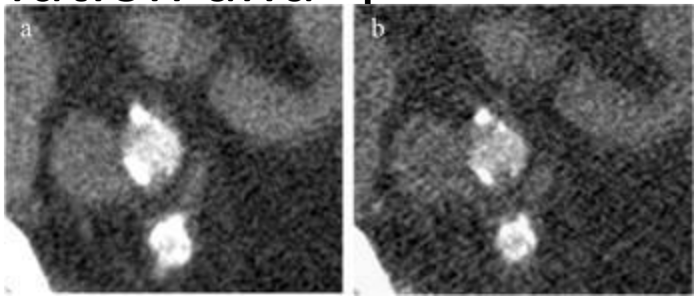
Number of detected photons depends on:

a) **mAs per rotation:** SNR α √mAs

b) **Slice width:** SNR α √slice width

↑slice width→ ↑ number of photons that contribute to each slice → ↓noise

But also: ↑slice width→ ↓ spatial resolution and ↑ partial volume effect



c) **kv:**

↑kv→ ↑ beam quality , and more photons are detected (even if mA is reduced to restore the dose)

Yet , this effect may be offset by the ↓ in contrast

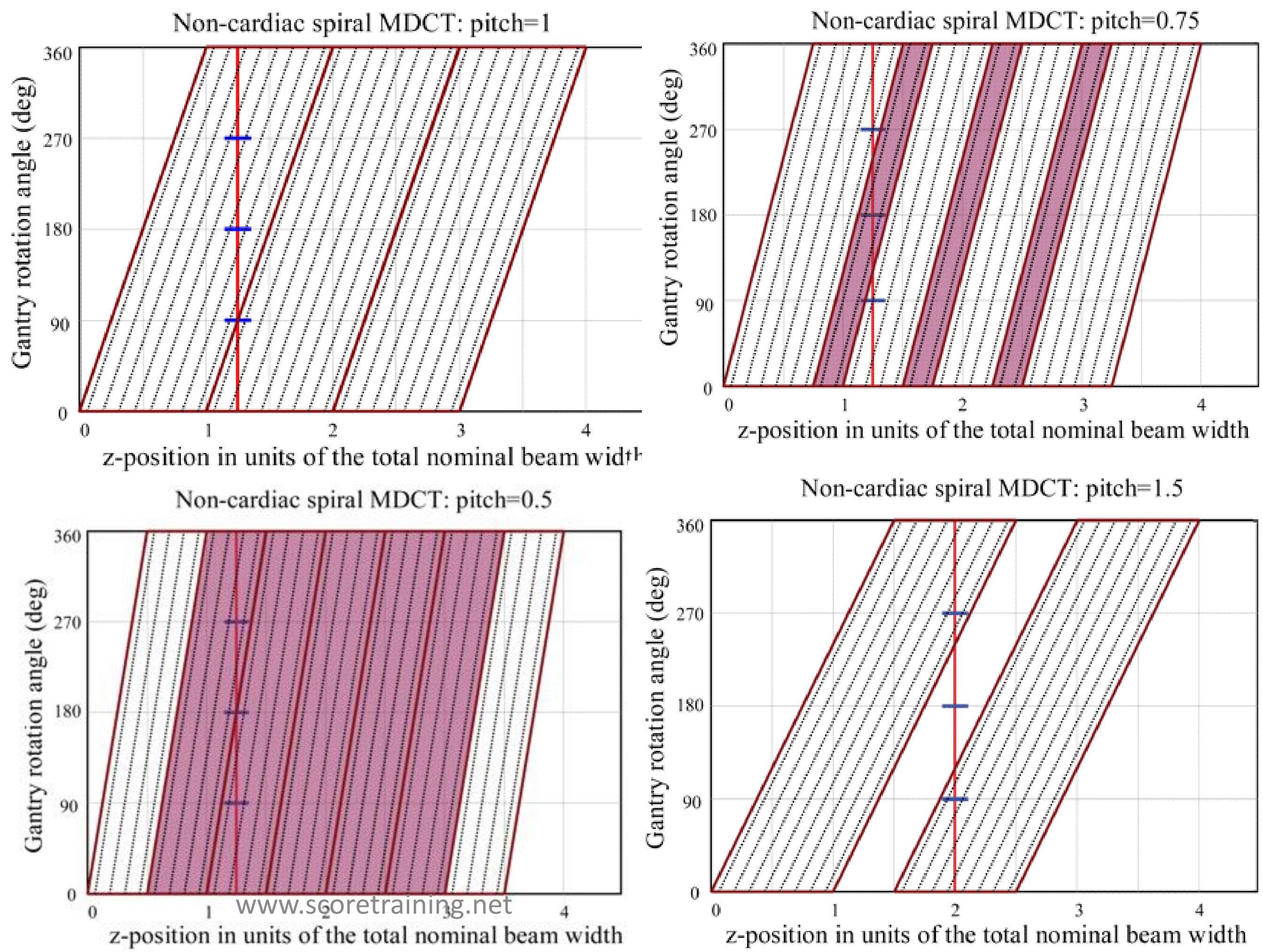
d) **FOV and pixel size:**

for smaller FOV and for larger matrix size → fewer photons per pixel → increased noise

e) **pitch:**

When pitch is less than 1: the measured spiral data partially overlap in the z direction (shaded areas) → more x-ray photons contributing to the images compared to when pitch = 1 → less image noise Consequently: When the pitch is less than 1 → noise is increased

N.B: The above is true only for multidetector scanner, The noise in single slice scanners are not dependant on the pitch



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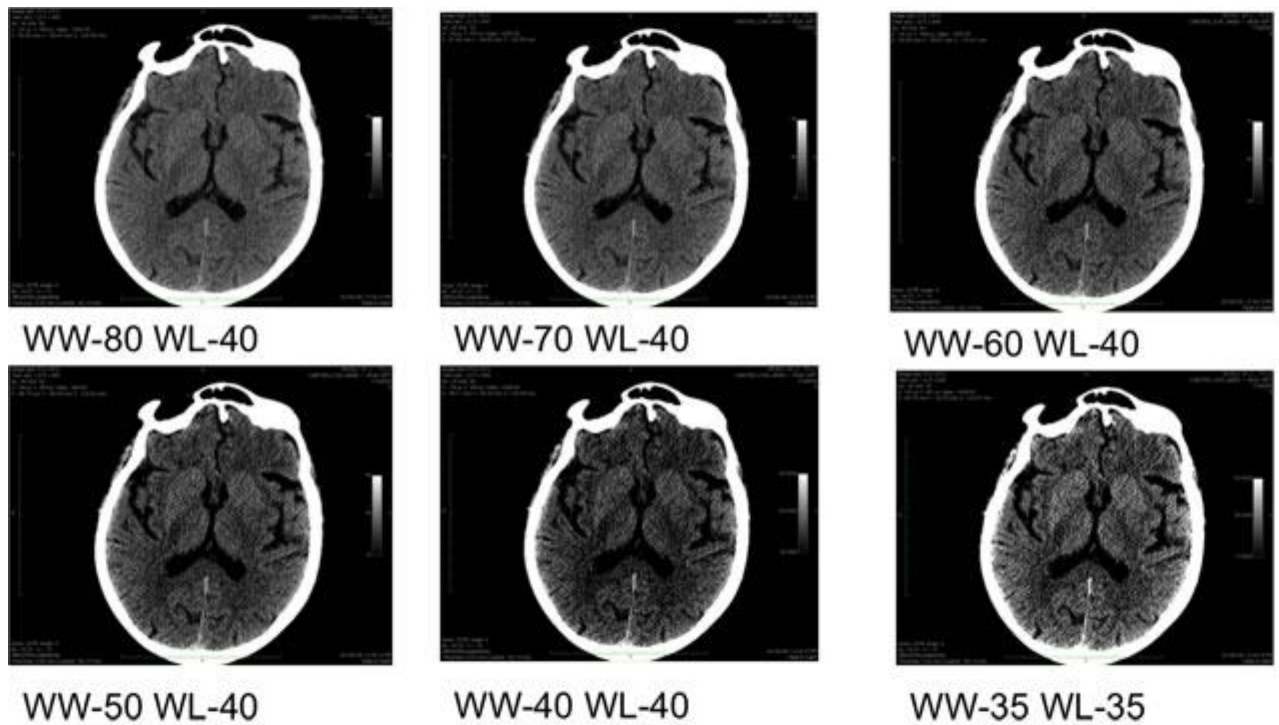
f) Window width:

- When window width is reduced → noise become more apparent

BUT: SNR remain unchanged



- If contrast between structures is comparable with the magnitude of noise →no amount of windowing will improve the ability to resolve the two structures

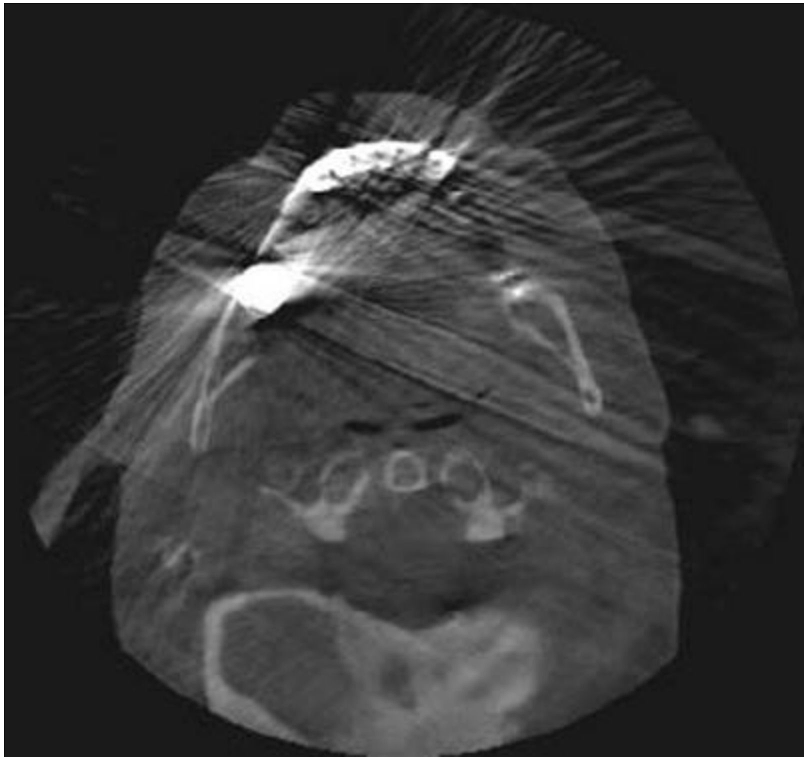
CT image artifacts

- **Beam Hardening artifact:**
- An x-ray beam is polyenergetic
- As the beam passes through an object, it becomes “harder” (its mean energy increases)
- Higher beam quality → decreased attenuation coefficient and CT numbers along the beam path (reconstruction assume that the x-ray is monoenergetic)
- this artifact is also called “cupping” artifact (resembles a cup)
- The hardening is most pronounced in the center and less at the periphery (beam is more filtered at the center of the field)
- Solution: bow-tie filters
- Common while imaging petrous bone

Motion artifact:

Causes: voluntary and involuntary motion

Result : the same structure occupy different voxels during the scan → streak artifacts



High attenuation (metallic) objects artifacts

Examples: metal implants , dental amalgam

Result: dark and white lines arising from the high attenuation material (streak artifacts)

Solution: metal correction algorithm (filtered backprojection)

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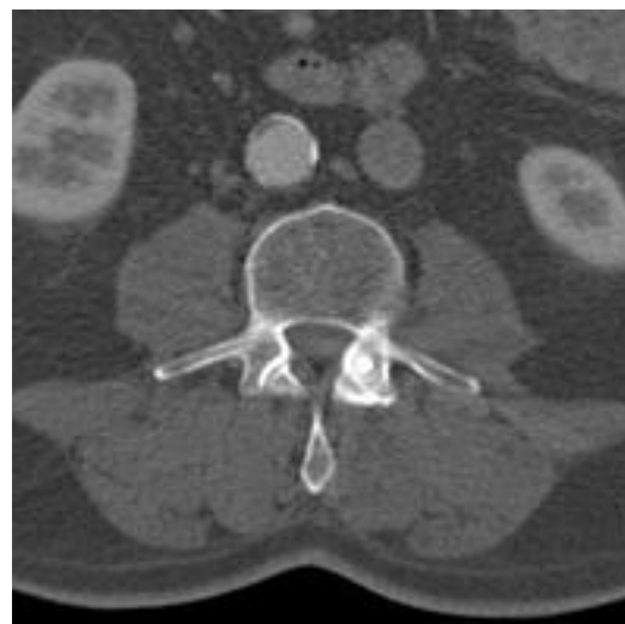

Photon Starvation

- **Cause**: highly attenuating areas such as the shoulders, pelvis with metallic implants .
- **Process**: When the x-ray beam is traveling horizontally, the attenuation is greatest and insufficient photons reach the detectors → very noisy projections are produced at these tube angulations. →horizontal streaks in the image (streak artifacts).
- **Solution**: mA modulation (see later)

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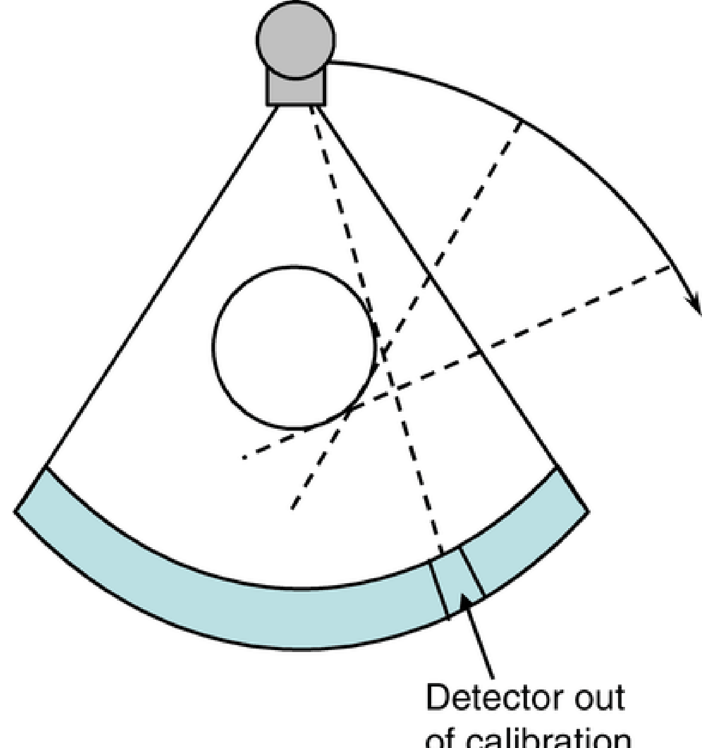
Ring artifact

- **Cause**: one of the detectors is out of calibration on a third-generation scanner.
- **process**: this detector will give a consistently wrong reading at each angular position, resulting in a circular artifact (light or dark ring)
- Larger changes in detector sensitivity is needed to make this artifact visible in multislice scanners compared to single slice scanners
- **Solution**: run automatic calibration programs when ct scanners are switched on every day



CT dose

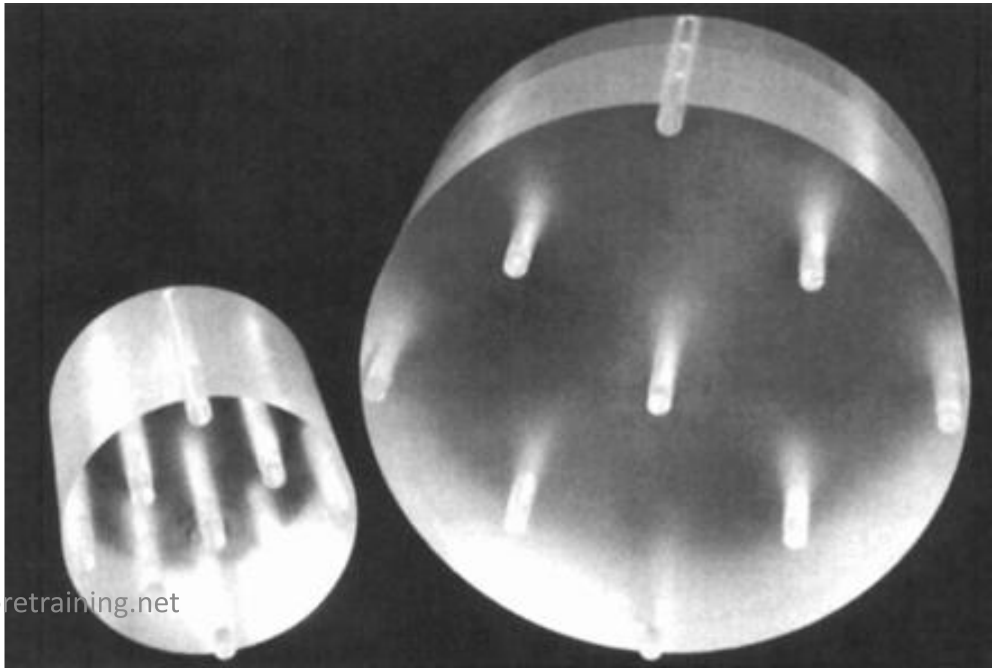
Two phantoms are used to measure patients dose:

a) a 32-cm-diameter cylindric acrylic phantom to represent an adult abdomen

b) a 16-cm-diameter version to represent an adult head or small pediatric bodies

Both are:

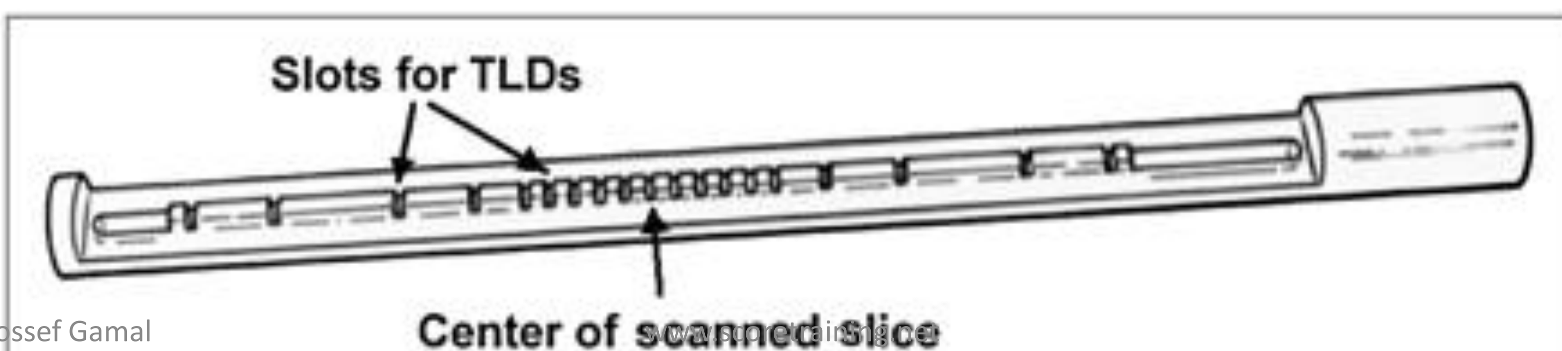
- 15 cm thick (in the *z*-axis direction)
- contain several 1-cm-diameter holes for insertion of dosimeters at the center of the phantom and at 3-, 6-, 9-, and 12-o'clock positions



Methods of measuring CT dose to single slice from the examination

1- using TLD (thermo-luminescent detectors)

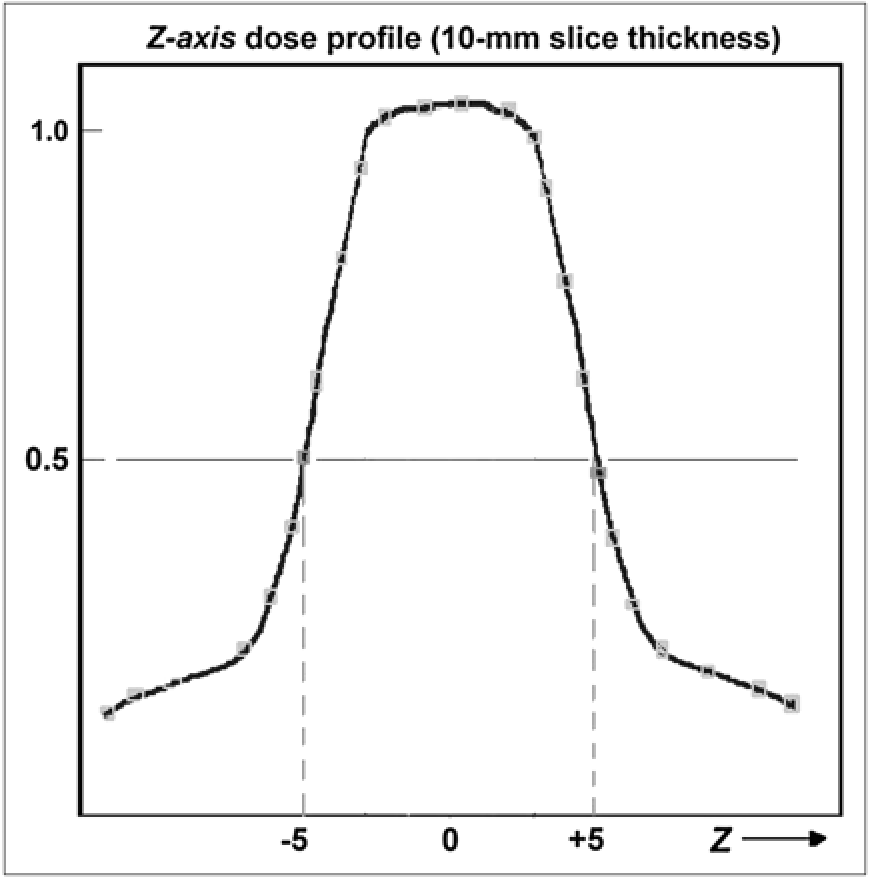
- TLD inserts are inserted in the phantom holes
- They designed to hold several TLDs, allowing closely spaced x-ray dose measurements within the x-ray beam and more widely spaced measurements outside the beam

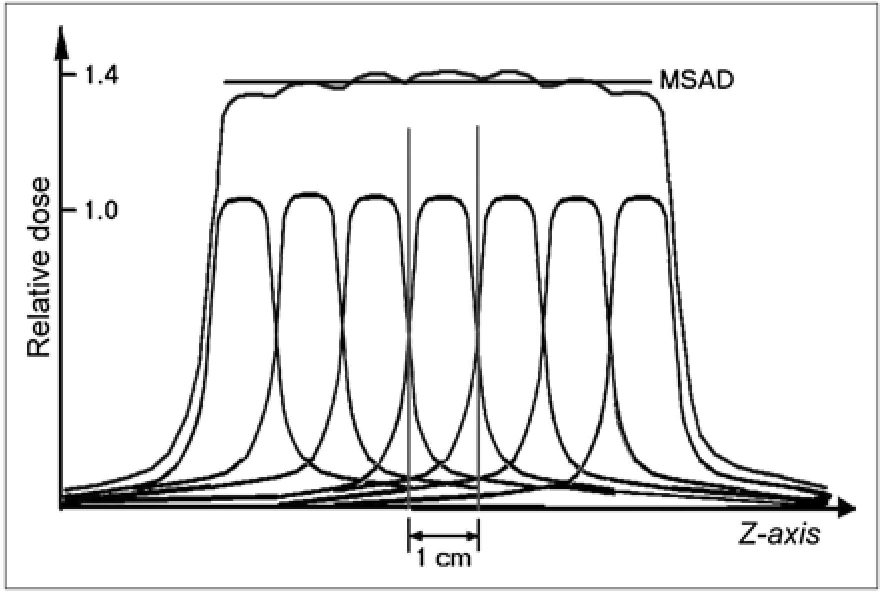


- a single slice scan is performed with the insert in place (centered on the slice).
- the radiation dose for each TLD is plotted against its *z*-axis position.

Two observations are noted:

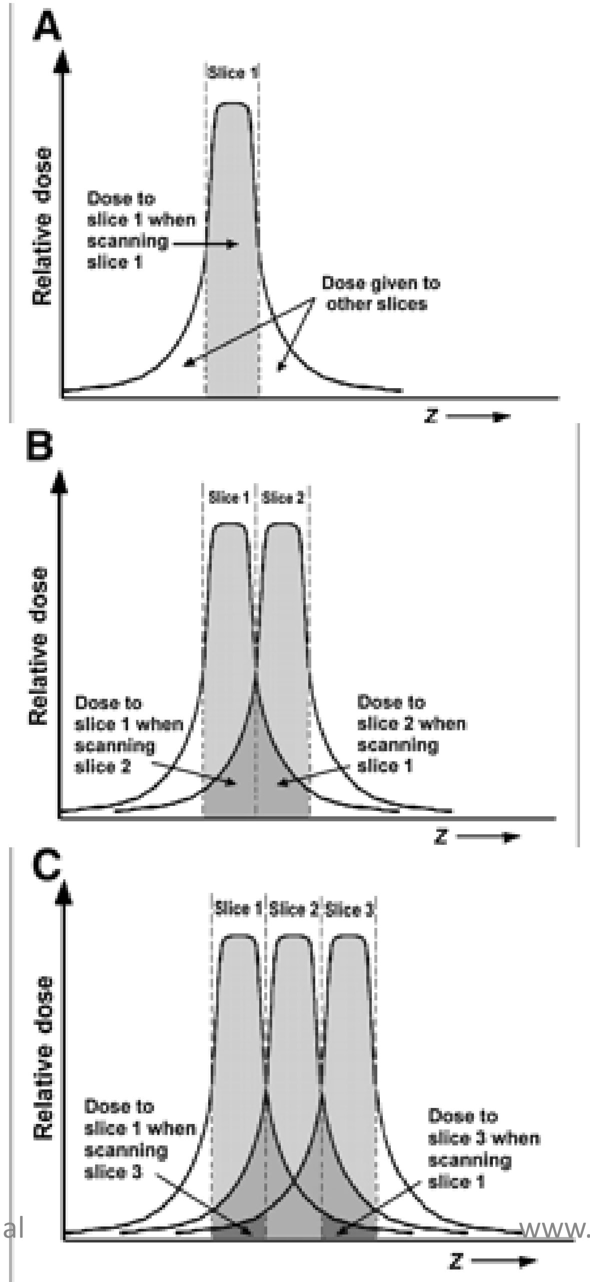
- 1- the profile width at the levels of 50% of the peak dose (full width at half maximum) is approximately the same as the slice thickness
- 2- a significant level of radiation still is outside the 10-mm-wide slice and deposits some of the dose into adjacent slices. (primarily from scatter)
- i.e. each slice of tissue receives radiation not only when that slice is scanned but also when adjacent slices are scanned.





- Cumulative dose from a series of contiguous slices are measured.
- multiple-slice average dose (MSAD): The average cumulative dose to the central slices received from the whole scan
- The MSAD may be 1.25–1.4 times the single-slice dose
- The cumulative dose to the end slices is lower than central slices because of lack of contribution from one side.

2- using single ionization chamber : Another method which directly measures a value closely related to MSAD



- A: single-slice profile when a slice 1 is scanned.
  - shaded region: dose that the tissue of slice 1 receives
  - tails of the dose profile: dose in adjacent slices.
- B: slice 2 is scanned.
  - The dose that the scanning of slice 1 gives to slice 2 equals the dose that slice 1 gets from the scanning of slice 2.
- C: scan a third slice
  - the dose that the scanning of slice 1 gives to slice 3 equals the dose that slice 1 gets from the scanning of slice 3.

- Conclusion: dose that scanning of slice 1 gives to all slices = dose that slice 1 gets from scanning of all slices
- So that to measure CT dose to single slice from the examination we can only measure the total dose under slice 1 profile
- This is achieved using ionization chamber long enough to cover all dose in tails of the profile
- The chamber is inserted into one of the phantom holes → single scan is obtained
- CTDI is measured:

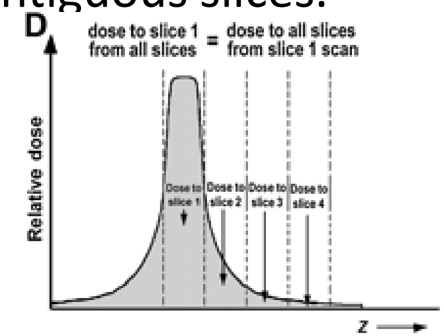
$CTDI_L$ (in rads or centigrays) = chamber reading x calibration factor x f-factor x L/T

CTDI: It is the dose to the phantom from a complete series of contiguous slices.

f-factor: (usually 0.87)

L: chamber length

T: slice thickness.



- for multislice CT : replacing T in the denominator with N × T, (= total *z*-direction beam width)
- N.B: CTDI and MSAD are equivalent; in fact, the only practical difference is the length of the dose profile included.

- CTDIs measured at different locations in a phantom will differ:

1- Doses for the 4 peripheral holes are nearly uniform

The CTDI measured at the 6-o'clock position somewhat lower because of table attenuation.

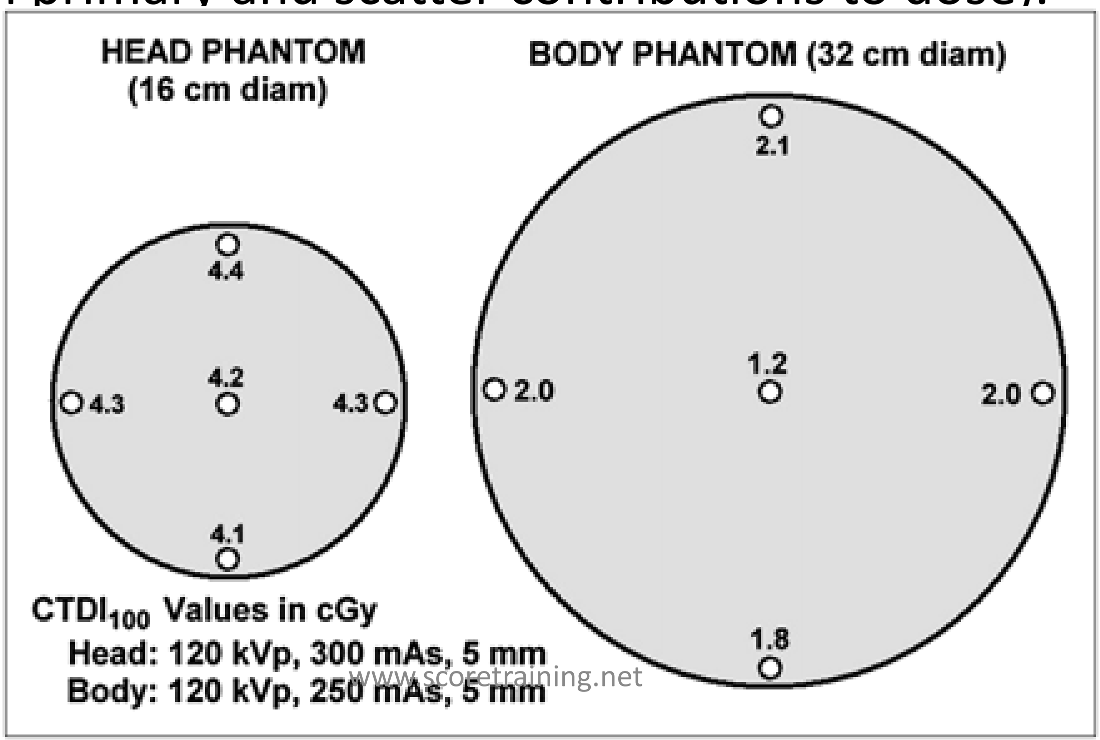
2- the central CTDI of the head phantom is nearly the same as that at the periphery

3- the central CTDI in body phantom is more than half the peripheral CTDI.

4- The dose from primary radiation is higher at the periphery

5- the dose from scatter increases significantly toward the center.

(CTDIs include both primary and scatter contributions to dose).



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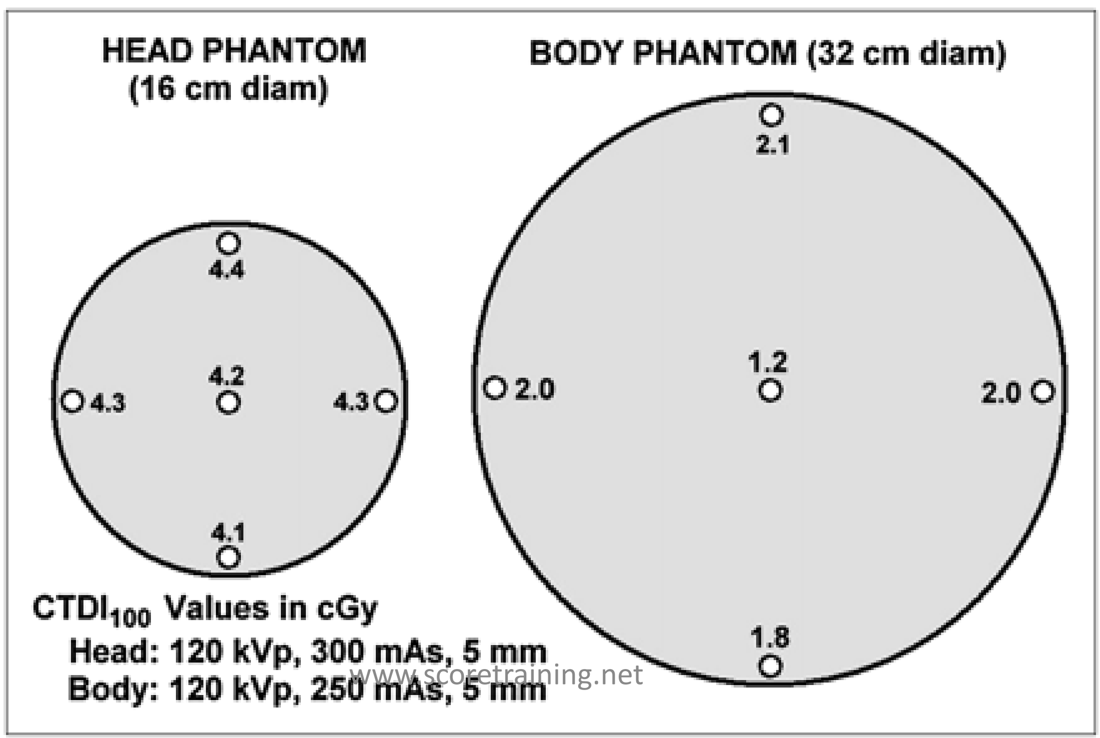
weighted CTDI ($CTDI_W$):

- Definition: average CTDI in the phantom,
- Equation: $CTDI_W = (2/3 \times CTDI_{periphery}) + (1/3 \times CTDI_{center})$

N.B: The CTDI measured at the 12-o'clock position is usually used for $CTDI_{periphery}$

N.B: $CTDI_W$ is often displayed on the CT operator's console during setting of the scan parameters.



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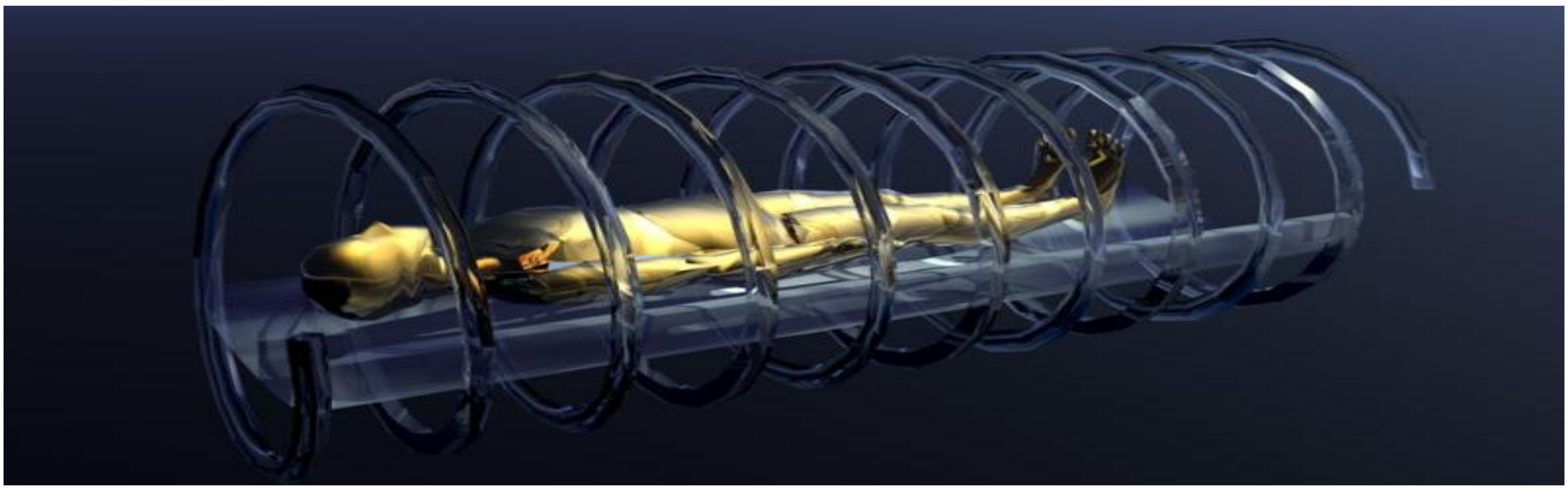
CTDI for PITCH ≠1 (Noncontiguous Slices)

$$CTDI_{volume} = CTDI_w / Pitch$$



Dose–Length Products

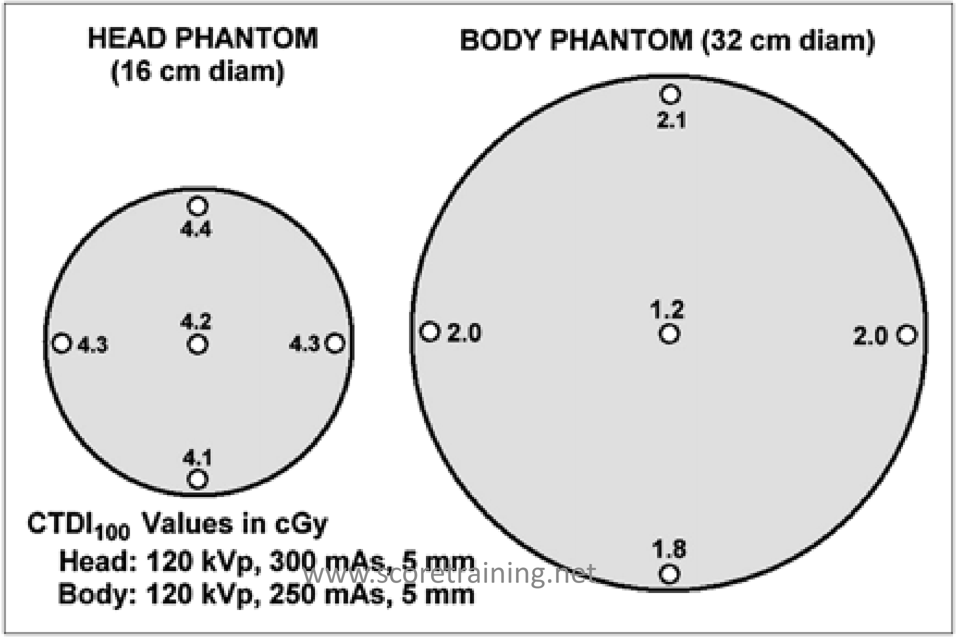
- Dose length product = L x $CTDI_{volume}$
- L: total *z*-direction length of the examination
- Dose length product indicates total amount of the radiation dose, while CTDI is an estimate of average radiation dose in the irradiated volume.
- Some CT scanners display dose–length product along with CTDI for each scan

Effective dose (D_E)

•**Definition**: the radiation dose that, if received by the entire body, provides the same radiation risk (i.e., of cancer) as does the dose received by the limited part of the body actually exposed

•**Equation:** D_E =DLP x conversion coefficient

•For head scans : weighting factor = 0.0023.
•For the abdomen, weighting factor = 0.17
Although the CTDIw for the head › body, the abdominal D_E is › head D_E (4 times) because the head scan irradiates a lesser amount of less radiosensitive tissue



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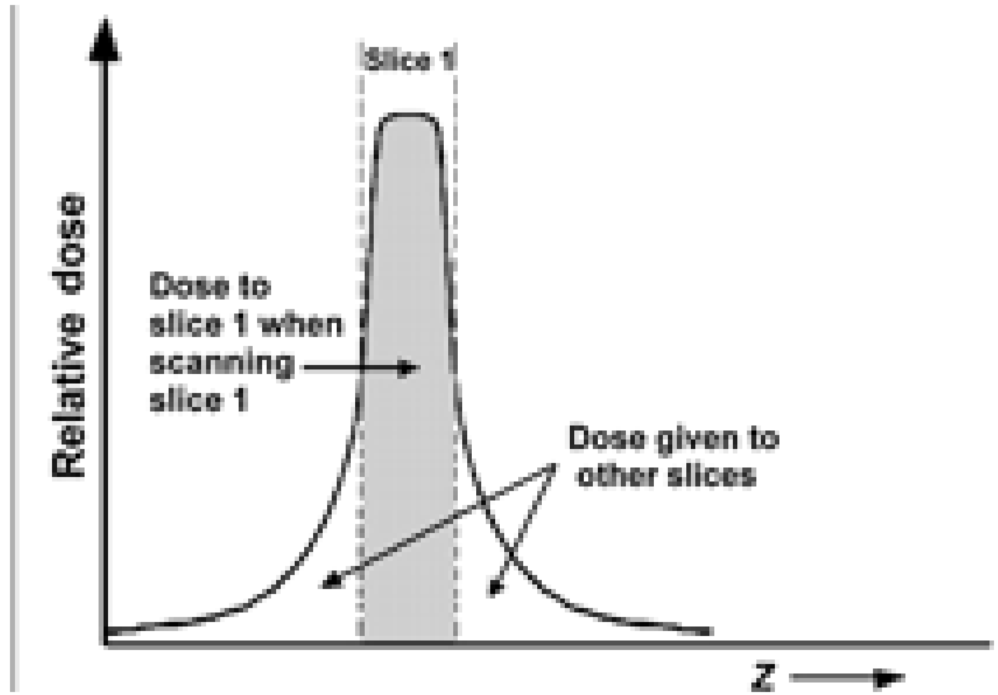
Maximum skin dose

- Equal to $CTDI_w$ For head scans , and 20% higher for body scans
- Skin dose of CT is higher than conventional radiography , but lower than fluoroscopy

Various Forms of CTDI

- $CTDI_{ideal}$: the measured dose profile length include the entire length of the profile (not practical)



- $CTDI_{regulatory}$: the measurement over a profile length corresponding to a specific number of contiguous slices (14 slices).
  - This quantity, must be reported by manufacturers in sales literature.
  - This definition will underestimate the dose for thin slices, e.g. a 3-mm thickness requires a CTDI measured over a length of 14 × 3, or 42 mm. (dose profile is significantly beyond those limits)

- $CTDI_{100}$:For practical purposes, a CTDI measured over 100 mm

Scanner Design Factors Affecting CT Radiation Dose

1- detectors geometry:

- Dose efficiency: fraction of primary x-rays exiting the patient that contribute to image
- has 2 components:
  - geometric efficiency: fraction of transmitted x-rays interacting with active detector areas
    - reduced if some x-rays are absorbed before detection (in detector housing) or pass between detectors
    - Geometric efficiency for modern third-generation single-slice scanners is relatively high (80%)
    - Geometric efficiency is reduced in multislice CT, due to the separations between detector elements in the *z*-direction create more dead space & because more of the *z*-direction diverted beam must be discarded
  - absorption efficiency: fraction of actually-captured x-rays interacting with active detector areas.
    - reduced if some x-rays that enter the detectors are not absorbed
    - absorption efficiency of Modern scanners with solid-state detectors: 99%.

2- distance of the x-ray tube from the detectors and isocenter

3-the design of beam collimator

4-The design of the bowtie filter.

Scanning Factors Affecting CT Radiation Dose

1)mAs : increasing mAs increases the dose proportionally

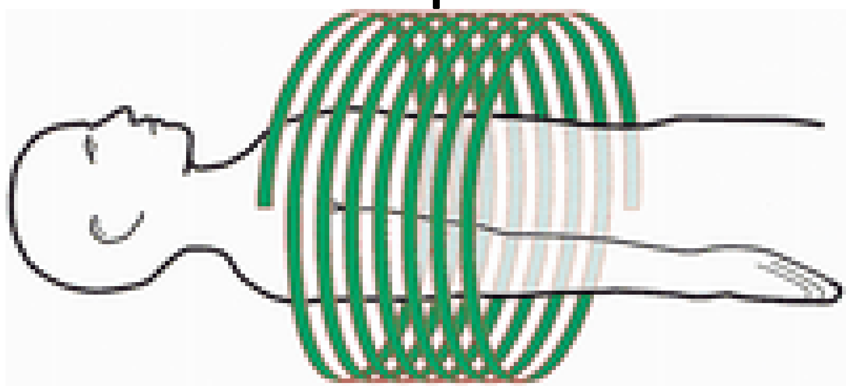
2) kilovoltage: a higher peak kilovoltage does not mean an increased patient dose and, in fact, allow the dose to be reduced.

3) slice thickness

- In single-slice CT with well-designed collimators:
- CTDI and DLP is independent of slice thickness for contiguous slices *($CTDI_L$ = chamber reading x calibration factor x f-factor x L/T)*
- Exceptions:
  - -in single slice scanners using post-patient collimators: beam size passing through the patient is greater than the slice thickness read →the narrower slice widths will increase the dose
  - - If mA is increased with narrower slices to keep the noise constant → increase of the dose

4) helical pitch

Patient dose (and CTDI) are inversely proportional to the pitch .



5)FOV: No direct effect on dose

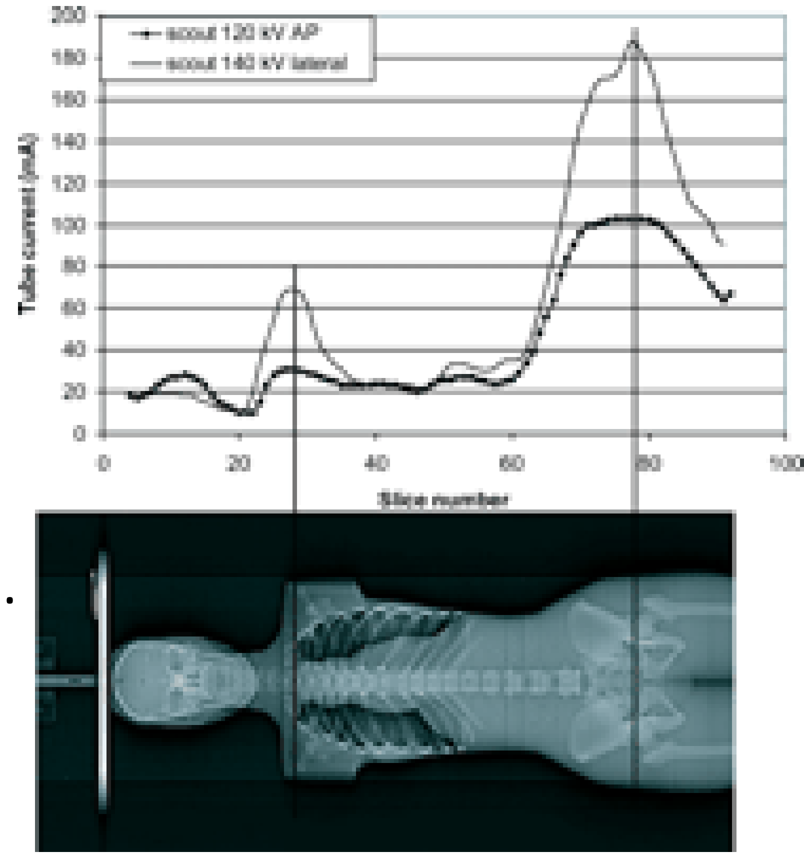
BUT may cause changes in mA to keep the noise constant

6)Type of the scanner

- Doses for helical scans are slightly higher than axial scans due to the addition gantry rotation at the start and the end of the scan
- Dose of multislice scanners is greater than single slice helical scanners:
  - A) greater over-scanning at the beginning and the end of the scan
  - B) over-collimation
- Number of detector raws used together in multislice scanners will not affect the dose

7) mA modulation:

- In past: a single mA value was specified (based on experience) for the entire scan length
  - → unnecessarily high mA values (and doses) for some slices,
- mA modulation
  - done Using information from an initial scout view the scan mA value is individually adjusted, depending on *z*-position, for each tube rotation.
- More advanced version of mA modulation:
  - mA adjustment not only for each rotation (*z*-position) but also as a function of angle during each rotation, e.g. pelvis anteroposterior and lateral thicknesses are different




Thank you.